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The Use of a Vehicle Acceleration Exposure Limit Model and a Finite Element Crash Test Dummy Model to Evaluate the Risk of Injuries During Orion Crew Module Landings

Charles Lawrence
Glenn Research Center, Cleveland, Ohio

Edwin L. Fasanella
Langley Research Center, Hampton, Virginia

Ala Tabiei
Multi-Physics Simulation Engineering, Mason, Ohio

James W. Brinkley
R&D Consultant, Kettering, Ohio

David M. Shemwell
Systems Planning and Analysis, Inc., Alexandria, Virginia

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National Aeronautics and
Space Administration

Glenn Research Center
Cleveland, Ohio 44135

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Edwin L. Fasanella
National Aeronautics and Space Administration
Langley Research Center
Hampton, Virginia 23681

Ala Tabiei
Multi-Physics Simulation Engineering
Mason, Ohio 45040

James W. Brinkley
R&D Consultant
Kettering, Ohio 45440

David M. Shemwell
Systems Planning and Analysis, Inc.
Alexandria, Virginia 22311

Abstract

A review of astronaut whole body impact tolerance is discussed for land or water landings of the next generation manned space capsule named Orion. LS-DYNA simulations of Orion capsule landings are performed to produce a low, moderate, and high probability of injury. The paper evaluates finite element (FE) seat and occupant simulations for assessing injury risk for the Orion crew and compares these simulations to whole body injury models commonly referred to as the Brinkley criteria. The FE seat and crash dummy models allow for varying the occupant restraint systems, cushion materials, side constraints, flailing of limbs, and detailed seat/occupant interactions to minimize landing injuries to the crew. The FE crash test dummies used in conjunction with the Brinkley criteria provides a useful set of tools for predicting potential crew injuries during vehicle landings.

Introduction and Background

The NASA Constellation Program plans are to replace the space shuttle orbiter with a Crew Exploration Vehicle (CEV) Command Module (CM) named “Orion.” Orion will be similar to Apollo but is larger and will carry a crew of four to six members while Apollo was designed for three astronauts. The Orion module will descend to Earth with a three-parachute recovery system in the event of emergency escape during the launch phase of the mission or during the final phase of the module’s return to Earth. The module is to be designed to be capable of a primary water landing as well as land landings. Landing systems must be designed for multiterrain impacts. Earth terrain is highly variable, with soil at proposed landing sites at Edwards Air Force Base in California typically very hard and soil at Carson Sink Nevada often very soft. During emergency escape from the launch pad, the module will be lifted away from the

launch rockets by a Launch Abort System (LAS) with a trajectory designed to land the module on water. Landing and recovery and protection of the crew members during water or land landings is a design challenge and despite years of successful space missions cannot be considered routine.

To keep the size of parachute systems reasonable and to optimize the weight of the landing system, the terminal velocity of the Orion module parachute recovery systems is to be designed to provide a descent velocity at landing of approximately 24 ft/s nominal to 33 ft/s off-nominal (16 to 22.5 mph/hr). In addition to the terminal descent (vertical) velocity, the parachute and module will move horizontally depending on the horizontal wind velocity. The velocities of the module at touch down are typical of crash impact velocities of small aircraft and helicopters. Thus, these impact velocities may produce injuries without some type of mitigation for impacts onto land. With proper pitch attitude into low sea states water landings do not require additional mitigation for nominal landings since the water acts as a natural impact attenuator. The Apollo capsule was designed to impact the water with a parachute hang angle of 27°. Since the Apollo entered the water pitched at an angle (thus giving it a wedge shape), the acceleration experienced was nominally 10 G or less for a 30 ft/s impact (Jones, 1970). The Apollo spacecraft also had crushable ribs and a crew seat pallet with stroking energy absorbers to alleviate off-nominal impacts. However, either a land landing of Apollo in an abort condition or a flat 0° attitude water impact with only one parachute fully opened was expected to result in occupant injuries. Since the probability of a pad abort with a land impact was considered to be low, the program accepted those risks.

Under ideal nominal conditions, either a land landing with retro-rockets or impact attenuating airbags or a water landing will result in tolerable accelerations to the crew if the seats and restraint systems are properly designed. For off-nominal landing conditions, a stroking crew seat pallet similar to the Apollo design is planned for the Orion module to reduce the impact accelerations to conditions that are tolerable to the astronauts and that will prevent injury. However, some mitigation for more severe off-nominal impacts can be designed with little weight penalty if a systems approach is taken early in the design phase.

The primary purpose of this paper is to review current whole body human impact tolerance models, including the Brinkley model, used by the Constellation Program for assessing injury risk during nominal and off-nominal landings of Orion. In addition, the paper will discuss the use of finite element (FE) seat and occupant simulations for assessing injury risk for Orion. These FE seat and crash dummy models allow for varying the occupant restraint systems, cushion materials, side constraints, flailing of limbs, and detailed seat/occupant interactions. The paper also includes a discussion of impact injury criteria used by the automotive industry to assess crash dummy response to impacts.

With properly designed seats with side supports and restraints, even relatively high accelerations can be tolerated with no or minimal injury, especially for impacts with the astronauts lying on their backs in a seat oriented as shown in figure 1. The local axis system has the +X-axis upward, the +Z-axis pointed toward the head, and the +Y-axis to the right. For accelerations along the +X-axis, which would occur for a flat impact, the acceleration in the spine-to-chest direction is commonly called “eyeballs in,” which reflects the inertial response to the acceleration. An acceleration along the –X-axis (chest-to-spine) is called “eyeballs out.” An acceleration from pelvis-to-head along the +Z-seat axis is called “eyeballs down,” and an acceleration pointing in the –Z-axis is “eyeballs up.” The seat coordinate system is a local system, and is a noninertial axis system. Hence, any angular rotations with respect to an inertial system must be considered.

Human tolerance to impact has been studied continuously since the advent of the automobile and airplane. The research that was accomplished to define the initial human tolerance to the +Z-axis accelerations associated with the development of emergency ejection seats in Germany, England, and the United States during and immediately after World War II was remarkable. Those accomplishments were followed by research efforts accomplished by others such as Shaw, Stapp, and Beeding to define tolerance limits for –Z, –X, and +X axes to support the development of ejection seats, both upward and downward seats, as well as crash tolerance limits for crew and passenger seats. Impact tolerance end points during this period ranged from spinal fractures to cardiovascular shock and retinal hemorrhage. In some cases tolerance levels were established on the basis of the judgment of the principal investigator.

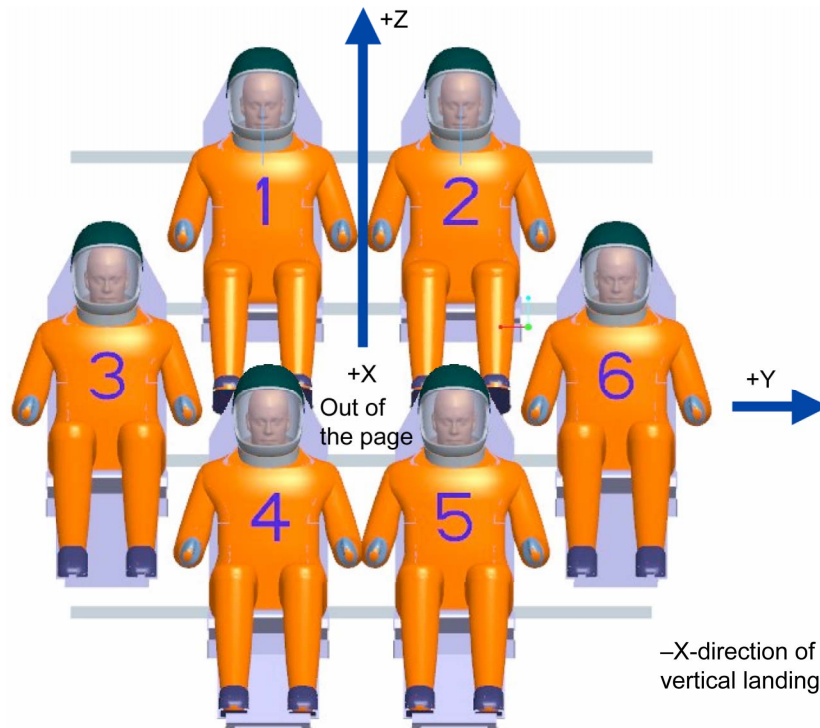


Figure 1.—Six-crew configuration and local X-, Y-, and Z-axes.

A comprehensive compilation of many of these early test results provides summary charts with regions of uninjured, moderate, and severe injuries versus maximum sled acceleration, rate of onset, and duration as well as protection system configuration was produced by Eiband at the NASA Lewis Research Center (1959). The data that Eiband used were limited to available experimental tests that had been conducted in the Z and X axes. Eiband concluded that adequate torso and extremity restraint is the primary variable in human tolerance to rapidly applied acceleration. When the conditions of adequate restraint are met, Eiband concluded that the other variables of acceleration direction, magnitude, and onset rate of the acceleration govern maximum tolerance and injury limits.

The impact tolerance charts produced by Eiband were useful; however, their application to an arbitrary acceleration pulse that includes accelerations along all three axes (X, Y, and Z) is quite subjective. An example of an Eiband chart for accelerations along the spine (+Z, eyeballs-down) is shown in figure 2. A modified chart (Snyder, 1973) with ejection seat design limits added is illustrated in figure 3. There are three areas to consider on each chart. The *Area of Voluntary Human Exposures* is the bottom-shaded area. In this region, all exposures were uninjured. For example, a maximum acceleration of approximately 14 G could be tolerated (uninjured) for up to 0.04 sec (40 ms). The other shaded area at the top of the chart is the *Area of Severe Injury*. For example, acceleration above 100 G for longer than 3 ms produces severe injury. The area between the two darkened areas is the *Area of Moderate Injury*. His analytical approach, which fitted experimental data with trapezoidal pulse shapes, has often been used to describe design acceleration limits. Unfortunately, that approach was found to be unworkable to analyze test data as acceleration profiles became more complex.

The advent of encapsulated seats for the B-58 bomber and Project Mercury, both having high +X accelerations and high rates of onset, led to experimental and analytical investigations of the complex high rate-of-onset acceleration waveforms (Payne, 1961; Holcomb, 1961; Headley et al., 1962) and of more robust protection systems such as rigid and semi-rigid, contoured body support systems (Brinkley, 1960). To support the development of the Project Apollo crew module, the Navy and Air Force conducted extensive studies of the influence of impact in multiple vector directions. Schulman et al., (1963)

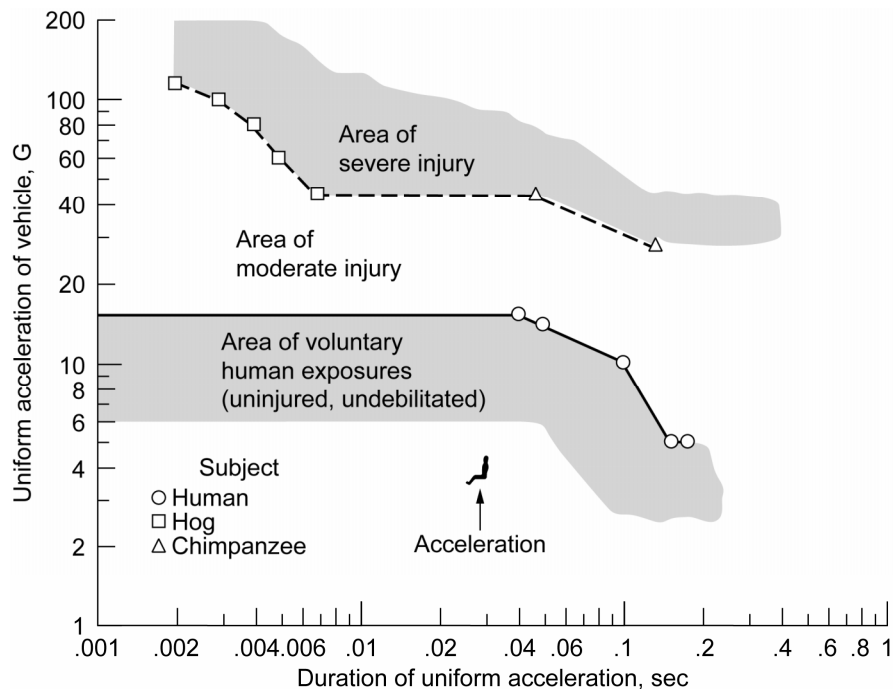


Figure 2.—+Z-acceleration tolerance with ejection seat data omitted (Eiband, A. et al.).

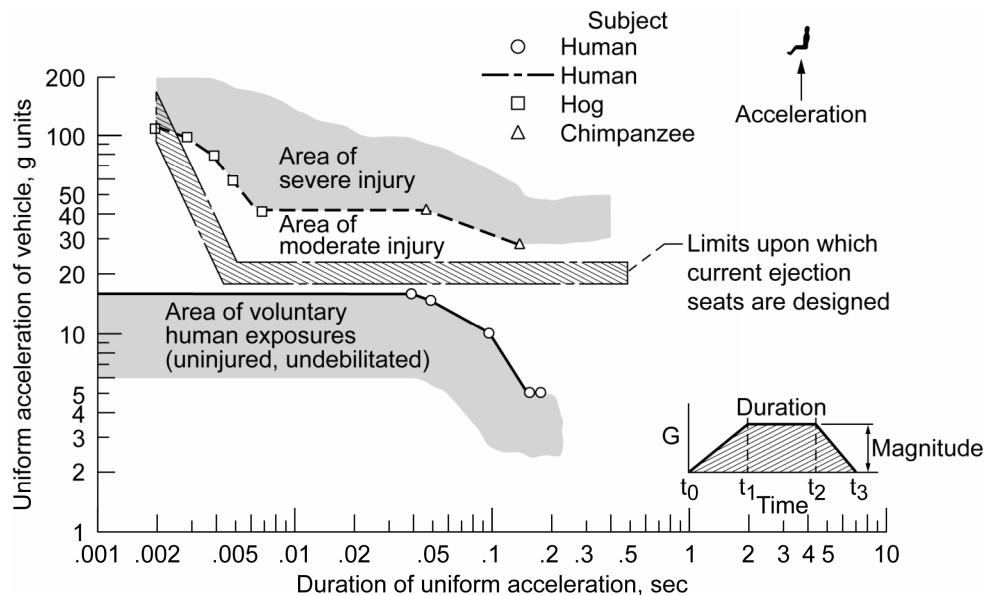


Figure 3.—+Z-human tolerance limits showing ejection seat design limits (Eiband, A. et al.).

conducted experiments to investigate -Z-axis impact levels up to 18.5 G with impact velocities up to 19.5 ft/sec. The volunteer subjects were restrained to a rigid couch by two shoulder straps, a cross-chest strap, lap belt, and crotch strap, and leg straps (fig. 4). Based upon observations of injuries suffered by cliff divers, spinal injury is probably the likely injury mode, but none of the volunteers experienced such injuries. Airmen using downward ejection seats have routinely been subjected to 10 G accelerations and a velocity change of approximately 22 ft/sec without spinal injury.

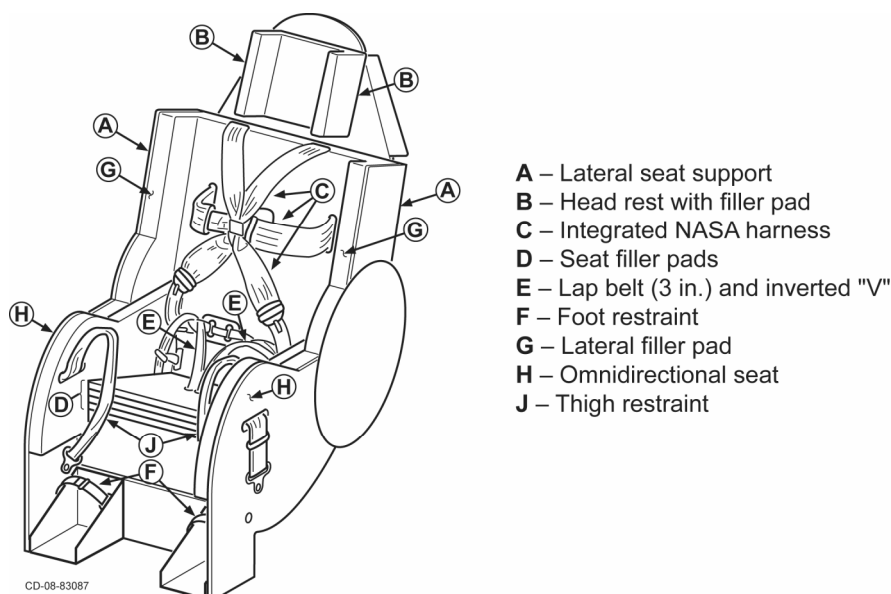
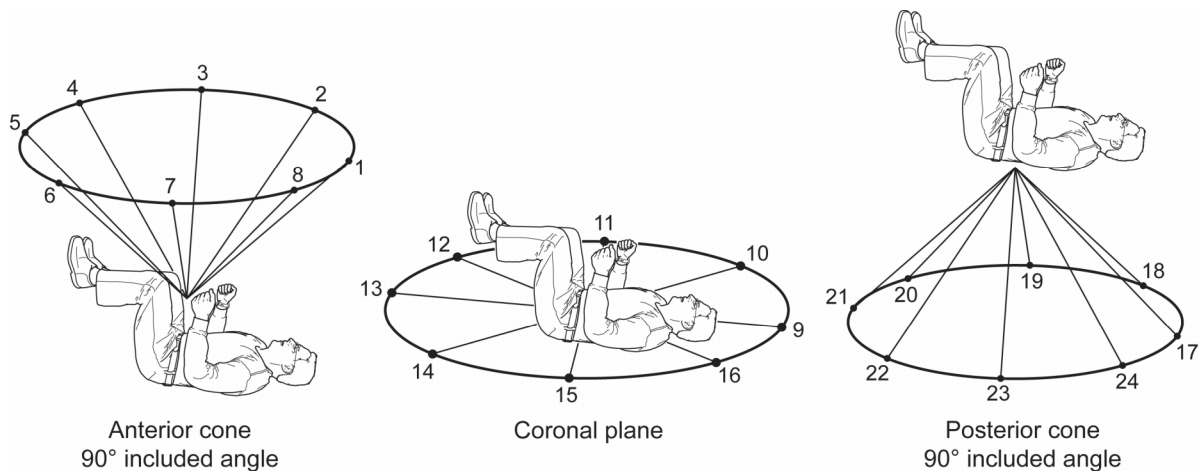


Figure 4.—Multi-directional impact test seat.

Weis et al. (1963) conducted 32 Y-axis vertical deceleration tests with 16 volunteer male subjects using a semi-rigid, contoured couch and restraint system at acceleration levels up to 21.6 G with impact velocities of 20.3 ft/s and rate of onset of 1350 G/s. A second series of 42 vertical decelerator tests were conducted with 25 male volunteers exposed to seven acceleration vector directions and six acceleration profiles. The subjects were restrained against a rigid body support system, which included metal side panels surrounding the subject's helmet, shoulders and hips. The subject's legs were restrained and their feet were braced within footrests. The acceleration vectors included: up 45°, up 45° right 45°, up 45° left 45°, right 45°, left 45°, left 90°, and right 90° (fig. 5). Peak accelerations were 26.0, 25.9, 25.5, 26.6, 25.4, 23.0, and 22.4 G, respectively. Impact velocities ranged up to 28 ft/s and maximum rates of onset ranged from 693 to 1270 G/sec at the maximum acceleration levels. Subject complaints and physiological measures did not indicate a tolerance endpoint had been reached during either test series.

During the Apollo program, Stapp and Taylor (1964) studied 146 impact tests with 58 male volunteer subjects conducted using a horizontal deceleration facility to predict the human response to estimated Apollo crew module landing conditions. A seat mounted within gimbals provided varied pitch and yaw so that the impact responses of the volunteers could be measured in 16 body positions with respect to the acceleration vector. Peak accelerations were 10, 15, 20, and 25 G. Onset rates were 1,000, 1,500, and 2,000 G/sec. Impact durations ranged from 0.60 to 0.130 sec. The subjects were restrained within a seat that provided sideward supports and restraint configuration similar to those used by Weis et al. All body positions and impact profiles were within voluntary tolerance limits except for the forward-facing, reclined position at 25.4 G with an onset rate of 1,000 G/sec and 0.060 sec duration in which compression of soft tissue around the 6th, 7th, and 8th thoracic vertebrae caused pain and stiffness for 60 days. Slowing of the heart (bradycardia) by as much as 90 beats per minute for 10 to 30 sec occurred immediately after impact in the -Z-axis direction and the acceleration exceeded 15 G. The authors recommended that the impact forces be kept below 20 G and above 0.060 sec pulse duration for spacecraft system design applications.

Brown et al. (1966) conducted over 288 impact tests for NASA with 79 male volunteers using the same impact facility, gimballed seat, and restraint system to evaluate the effects of 24 body positions. The acceleration onset rates ranged from 300 to 2500 G/sec. In general, lower onset rates were better tolerated. The impact forces caused effects to the nervous, cardio respiratory and musculoskeletal systems. Neurological effects were momentary stunning and disorientation. Transitory slowing of the heart rate was observed where the impact vector acted in the -Z-axis. Respiratory effects were momentary shortness of breath and chest pain. Musculoskeletal effects included spasms of muscle groups of the neck and back.



Sled positions (in degrees)																								
Position number	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24
Roll	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
Pitch	315	335	5	35	45	35	5	335	85	85	85	85	85	85	85	85	45	35	5	335	315	335	5	35
Yaw	0	330	320	330	0	030	040	030	180	220	270	320	0	040	090	140	180	210	220	210	180	120	140	150

Note: Deceleration force acts toward subject

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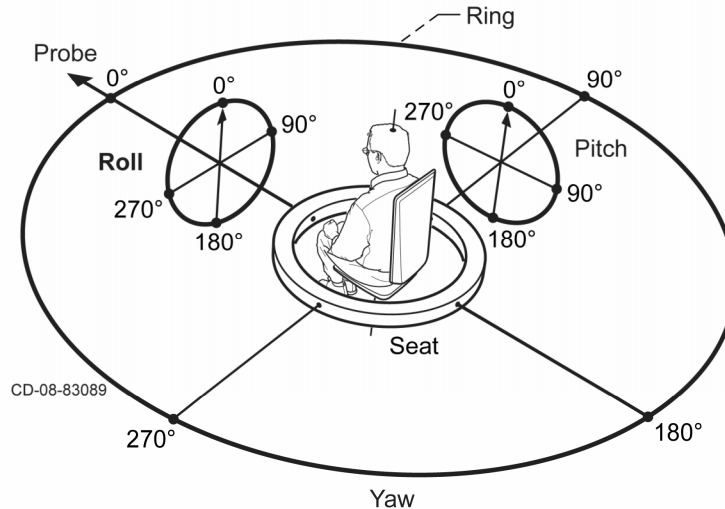


Figure 5.—Apollo impact test force vectors.

The results of these multi-axial impact tests apply to the Orion module where an emergency landing occurs and a moderate risk of injury is acceptable. Where Y-axis accelerations are high, the findings of these studies apply only if sideward body support is provided and the chest is restrained or supported in a similar manner. It is also important to note that a special restraint harness was used in each of the experiments conducted by the Air Force investigators (Weis et al., Stapp & Taylor, and Brown et al.). The restraint system was designed to provide extraordinary support for the upper torso, thereby reducing the load transmitted to the lower spine. In one case where the restraint was not adequately tightened, injury to soft tissues surrounding the 6th through 8th thoracic vertebrae occurred (Stapp & Taylor). Therefore, one

should not assume that the +Z-axis acceleration levels that were used could be tolerated using other restraint configurations.

Mathematical Lumped Parameter Injury Models

In order to reduce injuries produced during ejection from military aircraft, relatively simple but effective mathematical representations of a seated human were developed. Ejection seats were first designed based on acceleration limits in the +Z-direction and acceleration onset rates. Early on, it was determined that accelerations were better tolerated by aviators if the ejection acceleration onset rate was as low as practical. As ejection seats evolved and became more and more complex, such as those with encapsulated seats in the B-58, B-70 and the F-111 cockpit escape module, peak acceleration and onset rate proved ineffective for evaluating their performance especially during landing impact. New design tools became necessary. Payne (1965) and others suggested that relatively simple lumped parameter models could be used to represent the data that existed and could explain the relationships among acceleration input properties such as rise time, acceleration magnitude, and velocity change on the likelihood of injury. He also used such models to explain the effects of restraint system properties such as restraint slack and preload as well as seat cushion and impact attenuation system performance on the dynamic response of the human body.

The most effective human model for ejection seat design is the Dynamic Response Index (DRI) described by Stech and Payne in 1966. This work was performed for what is now called the Air Force Research Laboratory (AFRL) and was based on pioneering work in human biodynamics by von Gierke (1967) and his colleagues. The model was developed using data collected from vibration tests using volunteer subjects. The stiffness and breaking strength of the model were initially determined from tests of the vertebrae of cadavers. The overall frequency response was determined on the basis of the observed frequency response of the volunteers in laboratory experiments. The DRI model estimates of spinal injury rates were compared to actual operational ejection seat occupant spinal injury data (see fig. 6) by Brinkley (1968) and Brinkley and Shaffer (1971). On the basis of its validation using operation data, the Aeronautical Systems Center accredited the DRI model and incorporated it into the US Air Force specification used to develop escape systems and to qualify ejection catapults. In the mid 1970s, the model was accredited by the Air Standardization Coordination Committee and incorporated into an Air Standard ratified by air services of Australia, Canada, New Zealand, the United Kingdom, and the Air Force and Navy of the United States. The model was integrated into a computer program with limits for the other orthogonal axes, which were based upon then current estimates of acceleration and rate of onset limits. This program was used throughout the 1970s to evaluate the performance and to qualify US Air Force escape systems.

In the 1980s, Brinkley (1984) developed a method to estimate the likelihood of injury using lumped parameter models for each of the orthogonal axes. The effects of angular accelerations were included by incorporating the linear components of angular motion. The lumped parameter models were based upon available data from experiments conducted with volunteer subjects at both the US Air Force and Navy Biodynamics Laboratories as well as insights regarding injury modes developed from impact experiments using cadavers and animal subjects and more complex analytical models. The multiaxial dynamic response model was developed as part of a larger advanced development program to design and evaluate an ejection seat that would use sensors and a digital flight control system to control all of the components of an emergency escape system. This included a rocket thrust vector control system that could modify the acceleration magnitude and direction depending upon the initial airspeed, altitude, attitude, and life threat represented by these conditions. The method was successfully used within this program.

Further research was conducted to develop more accurate estimates of the model coefficients. Additional tests with volunteers and other available accident data were used to revise the model (Brinkley et al., 1990). The work of Salerno et al. (1989) is an example of an impact study conducted to measure the response of the human body and the head to impact in the +X axis. The revised model was used in several ejection seat test programs to evaluate their performance. After the method was used in the investigation

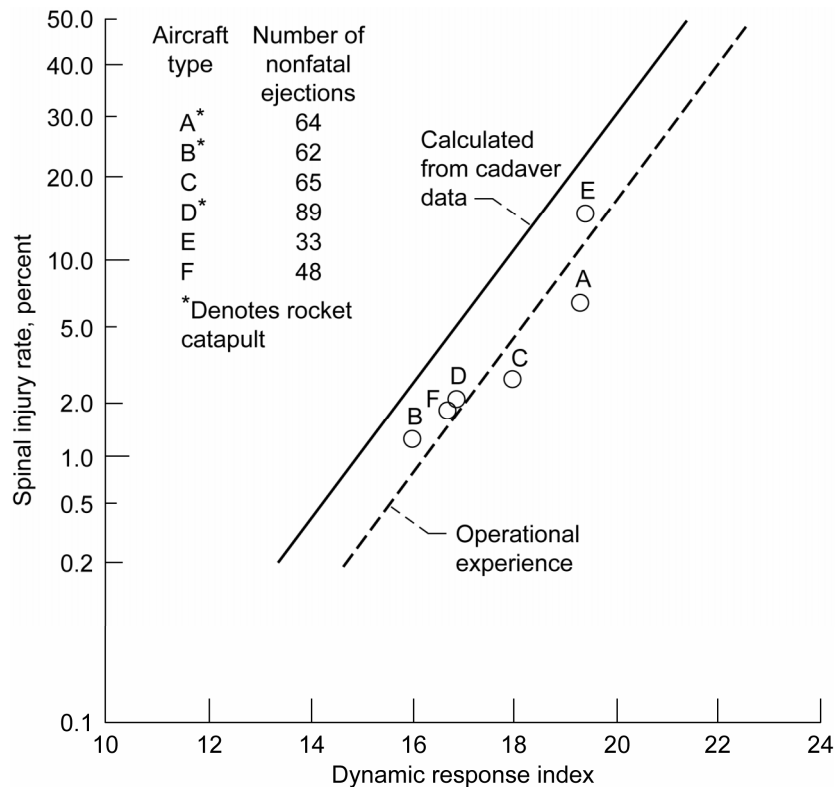


Figure 6.—DR +Z related to probability of spinal injury for ejection seats.

of the Challenger accident, medical personnel from NASA contacted James Brinkley and Lawrence Specker and requested further research to investigate the application of the model to the development of an emergency crew recovery system (The X-38) for the International Space Station. This work included use of the model to evaluate accelerations collected during landing of the Soyuz crew module and investigation of acceleration limits that would be appropriate for crew who were physiologically deconditioned due to prolonged exposure to microgravity, injury, and illness. The resulting effort focused on additional validation of the model using the Soyuz crew module impact data and on the potential effects of deconditioning due to microgravity. The results of this effort were provided to NASA Johnson Space Center (JSC) by briefings and the computer program used to accomplish the calculations. The revised model was used by JSC to design and evaluate the performance of the X-38 crew recovery vehicle.

Jennings et al. (2002) have summarized these physiological effects of exposure to microgravity as well as the efficacy of countermeasures. Predominate among these effects are orthostatic function during reentry and landing, and changes in bone and muscle strength. Many countermeasures have been attempted but have not been found to be fully protective. In view of these findings and the uncertainty about other factors such as astronaut age, gender, and other sources of physiological decrement that have not been quantified, the limit values were decreased for the landing impact event after reentry. Buhrman et al. (2002) independently reviewed the set of issues that could potentially degrade human impact tolerance. They recommended a reduction of the dynamic response values by an order of magnitude from the values used for emergency escape prior to space flight to account for the multiple risk factors that are involved. The authors concluded that the effects of combining multiple risk factors are unknown but could impose significantly greater injury risk than any individual factor.

The Multiaxial Dynamic Response Criteria

The impact exposure criteria that are included in the Brinkley model are similar in the technical approach to the models that have been developed to specify the human limits for human exposure to vibration and noise. These models are intended to specify the limits of the motion of vehicles at or near the position of the human occupant to prevent injury or performance decrements. The models and their associated limits are based upon whole-body responses or body organ responses to mechanical forces. The limits calculated from these models are often provided in simple curves or tables. For examples, see Nixon and Smith (2002).

The Brinkley model is a method that provides limits for vehicle accelerations occurring in any vector direction. The models for each axis represent the primary resonances that have been judged to be associated with the tolerance end points that have been reported in experiments with volunteers or observed in operational incidents. The method incorporates an approach that considers the influence of transient angular acceleration. The criteria that have been provided to NASA are based upon the presumption that the crew seats and restraint systems will be similar to the Apollo crew seats and restraint systems. Higher dynamic response limits may be appropriate if a more robust crew protection system such as used in the Soyuz crew module or such as those that were used to explore the response to multidirectional impact for Project Apollo cited earlier. A brief description of the Multiaxial Dynamic Response Method Criteria (referred to at NASA as the Brinkley Dynamic Response Model or Brinkley criteria) follows.

The equations of motion of the lumped parameter models that are used in the method are three second-order differential equations. Each equation is simply a forced spring-mass harmonic oscillator with damping. The forcing function “ A ” in equation (1) is the measured acceleration along each of the three seat axes. As a simplifying condition, the three equations are considered to be uncoupled.

$$\ddot{x} + 2\xi\omega_n\dot{x} + \omega_n^2x = A \quad (1)$$

Where:

\ddot{x} is the relative acceleration of the dynamic system in each of the X, Y, or Z axis.

\dot{x} is the relative velocity..

x is the relative deflection where a positive value represents compression.

ξ is the damping coefficient ratio.

ω_n is the undamped natural frequency of the dynamic system.

A is the component of the measured acceleration along the specified axis. Since the seat axis is not an inertial frame, rotational acceleration must be considered in terms of the linear components of the angular motion.

The dynamic response for each axis is given by:

$$DR = \omega_n^2 x / g \quad (2)$$

Where $DR(t)$ is the response of the dynamic system and g is the acceleration of gravity. Note that DR is dimensionless since it is divided by g , the acceleration of gravity. The value “ x ” when used along the +Z axis is the compression of the spine. Note that $DR(t)$ is essentially the normalized response of the occupant to the input acceleration in g . The maximum value of $DR(t)$ for any axis should be less than the limiting value. Thus, $DR(t)$ can be plotted versus the input acceleration time history. The maximum of the response $DR(t)$ can be greater or less than the maximum of the input acceleration $A(t)$ depending on whether the simple harmonic oscillator amplifies or attenuates the driving acceleration. The following values for ω_n and ξ shall be used:

$$\omega_{nx} = 62.8, \omega_{ny} = 58.0, \omega_{nz} = 52.9$$

$$\xi_x = 0.2, \xi_y = 0.09, \xi_z = 0.224$$

Limiting values of Dynamic Response, DR, levels have been set by NASA for Orion and they are listed in the table 1. These values are also provided in NASA CxP 70024. The very low injury risk (nominal) limits were derived by James Brinkley from studies initiated in the 1980s for NASA to accommodate a “deconditioned” astronaut who had been in space for a considerable time. The low injury risk (off nominal) DR limit is the same as is given in the report by Brinkley (1990). The limits in this report also consider DR levels for moderate and high risk of injury. This data is shown in table 2. Side restraints raise the dynamic response limits for the Y-direction as shown in table 2. Note that side restraints are not mentioned in the NASA HSIR specification. The limiting value of the Dynamic Response (DRlim) in each axis from table 1 is used in equation (3) to assess very low or low risk of injury. The value of beta in equation (3) must be less than one to pass the Brinkley criteria for very low or low risk. Note that beta could lag the input acceleration in time since the resonant frequencies of the lumped mass system are on the order of 10 Hz. Consequently, DR should be calculated until the maximum for each direction has been found.

$$\beta = \sqrt{\left(\frac{DR_x(t)}{DR_x^{lim}}\right)^2 + \left(\frac{DR_y(t)}{DR_y^{lim}}\right)^2 + \left(\frac{DR_z(t)}{DR_z^{lim}}\right)^2} < 1 \quad (3)$$

In the single +Z axis *DRI* model, the human body was modeled as a single lumped parameter model with a spring and damper representing the dynamic response of the human body for +Z-direction impacts.

TABLE 1.—BRINKLEY DYNAMIC RESPONSE LIMITS
[NASA HSIR Specs for Orion.]

DR level	X (eyeballs out, in)		Y (eyeballs right, left)		Z (eyeballs up, down)	
	DR _x <0	DR _x >0	DR _y <0	DR _y >0	DR _z <0	DR _z >0
Very low (nominal)	-22.4	31	-11.8	11.8	-11	13.1
Low (off-nominal)	-28	35	-14	14	-13.4	15.2

TABLE 2.—DYNAMIC RESPONSE LIMITS FROM BRINKLEY,
J.W., SPECKER, L.J., AND MOSHER, S.E.

DR level	X (eyeballs out, in)		Y (eyeballs right, left)		Z (eyeballs up, down)	
	DR _x <0	DR _x >0	DR _y <0	DR _y >0	DR _z <0	DR _z >0
Low (same as NASA spec)	-28	35	-14	14	-13.4	15.2
Moderate	-35	40	-17	17	-16.5	18
High risk	-46	46	-22	22	-20.4	22.4

Note that if side restraints are used, the DR levels for Y increase to -15 and 15 (low), -20 and 20 (moderate) and -30 and 30 (high risk). Caution—these values are not in the NASA HSIR—use table 1.

Thus $DR_z(t)$ is basically the nondimensionalized response of the lumped parameter model of the torso to a forced acceleration input along the Z-axis. In general for multiaxis accelerations, the linear second-order differential damped oscillator given by equation (1) is solved independently for each direction. The input acceleration for the model should ideally be the acceleration of the seat.

From operational data, the Dynamic Response Index (*DRI* for +Z-direction from eq. (2)) was correlated to injury using the data shown in figure 6. From the chart, most of the DR data points from operational experience ranged from a low of 16 to a high of 20. Any DR greater than 22 or 23 was correlated to a spinal injury rate above 50 percent. Note that the DR+z limit (DR_{+z}^{lim}) in equation (3), shown in table 1 for off nominal in the +Z direction is 15.2, which corresponds to an injury probability of less than 1 percent. The DR+z nominal from table 1 is 13.1.

Although the Brinkley model has had success in predicting injury, it is a simplified model of a very complex dynamic system (the human body). Also, the method of combining the results of accelerations in the X, Y, and Z-directions is not fully verified as most test data were gathered for a single acceleration direction. The influence of acceleration vectors between the primary axes has been explored by Weis et al., Stapp & Taylor, and Brown et al. Insight is obtained by looking at the DR for each direction before combining the results to obtain Beta in equation (3). Finally, note that the Brinkley model is only used as a tool for estimating the likelihood of injury. The model was not developed to predict the exact nature or severity of the injury. The criteria were not correlated with fatalities nor were they intended to be. The threshold for fatalities is very difficult to predict. When the Brinkley criteria were provided to NASA JSC in 2004 during the development several assumptions were made. First, it was presumed that the body support and restraint system would be similar to the system used in the Apollo crew module. This presumption leads to the conclusion that there would be minimal side support. However, increased DR values were provided as an incentive for the use of side supports although it was also presumed they would be minimal. From discussion during the meeting when racing type seats were considered, it seemed unlikely that large side supports such as are used in racing or used during the Air Force Apollo test programs would be acceptable in view of astronaut seat layout limitations, workspace, weight, and emergency crew egress requirements. Second, the restraint system is presumed to be adequately pretensioned to eliminate slack. Third, the +Z-axis limits assume the seat cushion materials do not amplify the acceleration transmitted to the seat occupant. Fourth, the +X-axis limits presume the seat occupant's head is protected by a flight helmet with a liner, which is adequate to pass the impact test requirements of ANSI Z-90 (latest edition) or equivalent specification. Fifth, a caveat was provided which read: The dynamic response models that are used in the method are descriptive of available empirical, whole-body response data, but the models (except for the +Z-axis model) do not attempt to predict the exact mode of injury, i.e., compression fracture of the lumbar and thoracic vertebrae. The risk of specific modes of injury, such as head injury, should be evaluated using other methods. Salerno et al. (1987) was referenced as a potential source of information on the dynamic response of bare head to impact and head impact injury although many other sources are available.

The Brinkley model is limited in that it uses a simplified mechanical model of the dynamic whole-body response of the human. A schematic of the physical layout of the Brinkley model, showing the driving point for the model located in the mid-thorax, is shown in figure 7. The nature of the Brinkley model and its representation of the human body as a lumped parameter model acting at a single point preclude the model from distinguishing relative movement of individual body parts. While this model works reasonably well for vertical Z-axis loads, such as those of an ejection seat, where the major injury location is in the lower spine and the human body moves as a unit, it does not properly physically model some other impacts, such as side impacts, where sections of the body like the head relative to the body center of mass. In addition to the Brinkley model not having the capability to discern different types of injuries, the Brinkley model cannot provide insight into the effect of any advanced restraint system that provides additional restrictions or protection to the head or other parts of the body.

In summary, the Air Force Research Laboratory developed human tolerance models that were grounded in models to assist in the design and evaluation of emergency ejection seats, escape capsules, and spacecraft. Data collected from impact tests using military volunteers, the results of operational

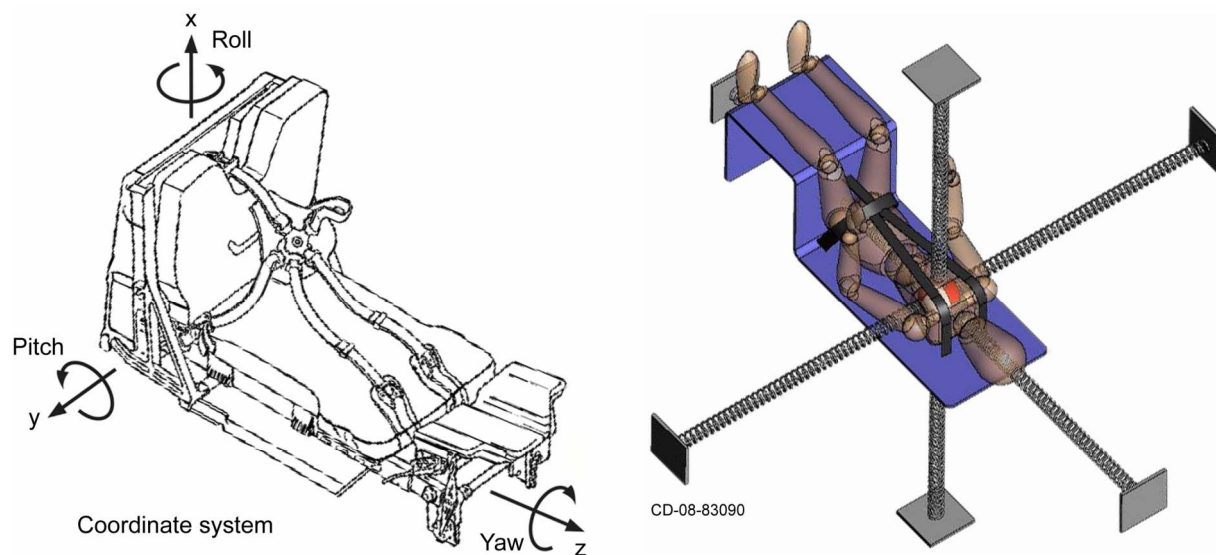


Figure 7.—Brinkley single degree of freedom body model.

escape system and space vehicle experience, expert judgments founded on observations of responses of volunteers to non-injurious acceleration conditions, available impact injury data, and insights gained from computational simulations of human responses were used to develop and validate the model. The model has proved to be extremely useful and is included in the military specifications for aircraft crew escape systems and has been incorporated into NASA Standard 3000. The model and its impact exposure criteria have been used to evaluate the performance of emergency escape systems including the ACES II, the Russian K-36D ejection seat, several developmental ejection seats, and is currently being used to evaluate the ejection seat for the US Air Force F-35 aircraft and developmental ACES V ejection seat. The model has also been used to evaluate the X-38 landing impact and the landing impact of several Soyuz crew modules.

Other Impact Acceleration Criteria

In contrast to the Air Force, the Army developed design guides requiring features such as crashworthy cockpits and seats to protect the crew in the event of a “survivable crash” (Coltman et al., 1989). The automobile industry was faced with federal regulations to design and build safer cars. Anthropomorphic Test Dummies (ATDs) with more human impact fidelity such as the Hybrid III were developed as instrumented passengers for controlled crash tests. The National Highway Traffic Safety Administration (NHTSA) regulates automobile safety and also sponsors research to make automobiles safer. A description of many of the injury criteria used in the automotive industry is provided in Schmitt, K. -U., et al.

The current state of the art does not allow the direct modeling and simulation of a human body subjected to injurious high acceleration loading or impact conditions. To address this limitation the automotive industry has attempted to generate a relationship between the physical behavior of the various crash test dummies and the human body. This has been attempted via a combination of cadaver, sub-injury level testing on human volunteers, animal experimentation and data gathering from actual accidents. While the automotive industry has years of experience correlating test crash dummy behavior during controlled automobile crashes with actual real world crashes and human injury little work has been done to correlate the dummy behavior with the impacts and accelerations expected to occur during Orion landings.

The general mechanical difficulty in developing a single three-direction anthropomorphic crash test dummy that has adequate levels of bio-fidelity has proven to be challenging and has motivated the automotive industry to develop a series of crash test dummies specific to various types of automobile impact conditions. The three main types of anthropomorphic crash test dummy types of interest are: frontal collision full body motion, rear impact, and side impact. For frontal impacts the industry typically utilizes the Hybrid III anthropomorphic crash test dummy as the standard. This crash test dummy has been

calibrated with years of test data and is considered a good indicator for whole body motion for frontal impact. The HYBRID III dummy, however, is considered to be inadequate for rear impacts or for impacts in the spinal directions. The automotive industry has developed a number of rear impact anthropomorphic crash test dummies. The rear impact crash test dummies have more accurate head and neck mechanisms than the Hybrid III crash test dummy and are better able to predict neck and whiplash types of injuries. The rear impact dummy, (RID), series of crash test dummies include RID1, bio-RID, RID2 and Bio-RIDII. The most important aspects of these anthropomorphic crash test dummies are the additional attention to the mobility of the spine particularly in the neck region and also in the area near the hips. The third type of dummy is the side impact dummy series designated SID. The SID's are similar to the RID's but with additional fidelity for spinal motion in the sideways, y-axis, direction. To complement physical crash testing the industry has developed finite element models of the various anthropomorphic crash test dummies.

In the present effort the Hybrid III FEA model was selected as the baseline. This is a model of the most commonly used crash test dummies and was thought to be a good starting point for comparing the results from the crash test dummies to those from the Brinkley model. It was also thought that the Hybrid III was a reasonable crash test dummy for assessing alternate crew protection concepts. It is recognized that for situations such as side and rear impact models such as the Side Impact Dummy (SID) or Rear Impact Dummy (RID) are more suitable, however, the incorporation of these models is left for future work.

In order to directly relate the results of the FEA simulations, which provide forces and moments in strategic locations throughout the model, and actual injury, the nature of specific relevant injury and the forces or moments that produce it must be understood. Both the Air Force and the Navy have active programs in the area of biomechanics and injury potential of ejection seats. Much of this work is summarized in Bowman, B.M. (1993). This work was commissioned by the Navy as part of an effort to better understand the use of anthropomorphic crash test dummies in the evaluation of escape systems. This paper is a survey of the state of the art in understanding the effect of ejection seat G forces on the human body. In particular the data indicates that injuries that are experienced at these acceleration levels (plus or minus Z axis only) are almost always related to injuries of one type or another to the spinal column. Serious injuries to soft tissue that are not directly related to impact are effectively nonexistent. This fact is important to the CEV effort since it indicates that the focus, at least in the Z axis direction, can be primarily on predicting and preventing spinal column injuries.

Finite Element Crash Test Dummy Simulations

The Orion (version 604) Crew Module (fig. 8) was used to generate vehicle responses so that a comparison could be made between injury predictions from the Brinkley model and predictions from the FE crash test dummy. To generate the response, the vehicle was oriented in three primary human body directions; eyes in/out (z axis), sideways (y axis) and spinal (z direction). The vehicle was given an initial velocity just before impact with the ground. These orientation exaggerate realistic landing scenarios, however, they provide extreme cases for comparing the Brinkley model to the crash test dummy results. The finite element program, LS-DYNA was used to perform the analysis of the vehicle impacting ground and resulting acceleration profiles were computed and extracted at the location of the center of mass of the crew members. The extracted acceleration profiles were then input into the Brinkley model and the level of injury was computed using the Brinkley injury criteria. The initial impact velocity was adjusted until the velocities corresponding to high, medium and low injury risk were identified (table 3). Once the impact velocities corresponding to the three levels of injury risk associated with the Brinkley model were determined, and the acceleration profiles generated (figs. 9 to 20) a separate LS-DYNA model was created that included only the FE crash test dummy constrained in a seat with a five point harness. The acceleration profiles were then applied to the seat in this model, simulations were run, and a comparison was made between the Brinkley injury criteria and the response extracted from the FE crash test dummy. It should be noted that accelerations resulting from the effect of gravity are included in all of the simulations performed in this paper.

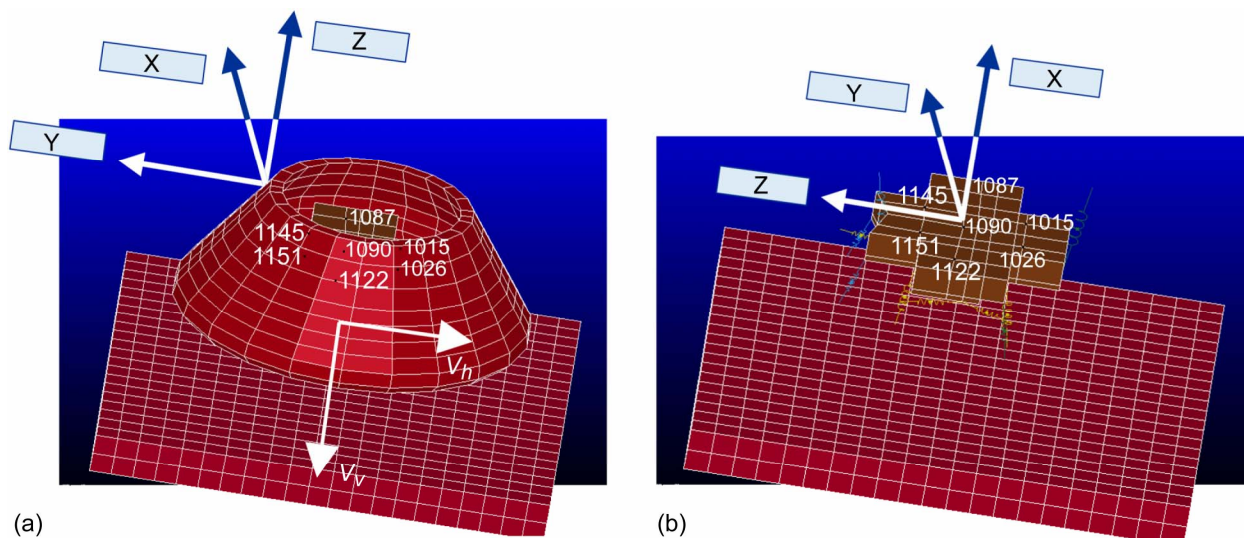


Figure 8.—The 604 seat attenuation model. (a) Global coordinate system. (b) Brinkley local coordinate system (pressure vessel not shown).

TABLE 3.—VERTICAL VELOCITY AND BRINKLEY INJURY
[Landing vertical velocity (fps)/Brinkley Dynamic
Response Index (DRI).]

	Eyes in/out	Spinal	Sideways
Low injury risk	33 fps/29	25 fps/13	17 fps/9.2
Medium injury risk	42 fps/35	29 fps/15.8	25 fps/17.4
High injury risk	58 fps/43.9	33 fps/19.2	33 fps/23.2

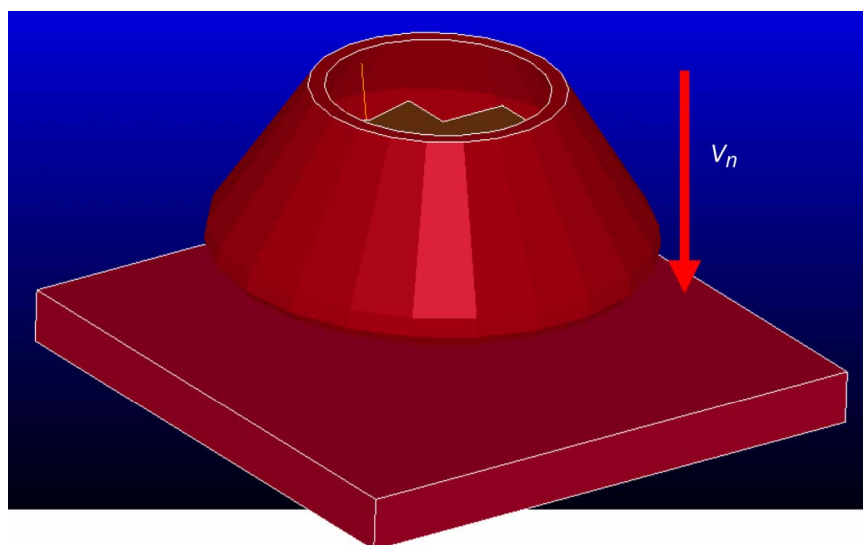


Figure 9.—Configuration for X-acceleration (eyes in/out DR).

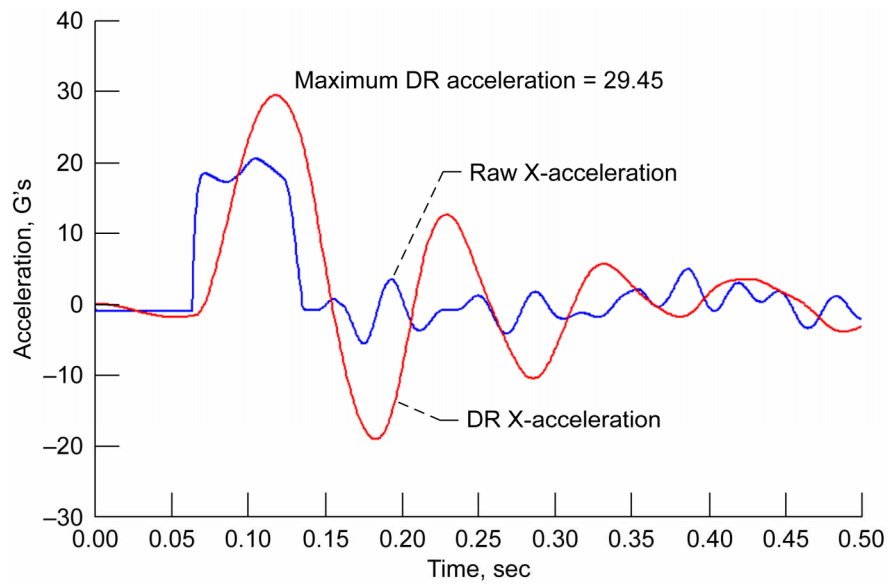


Figure 10.—Acceleration profile for low eyes in/out DR ($V_n = 33.3$ fps).

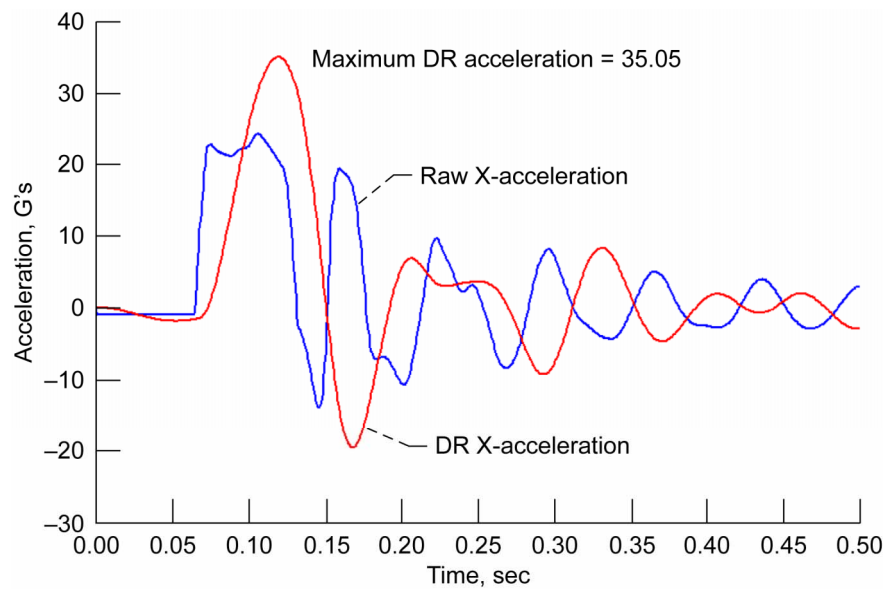


Figure 11.—Acceleration profile for moderate eyes in/out DR ($V_n = 41.67$ fps).

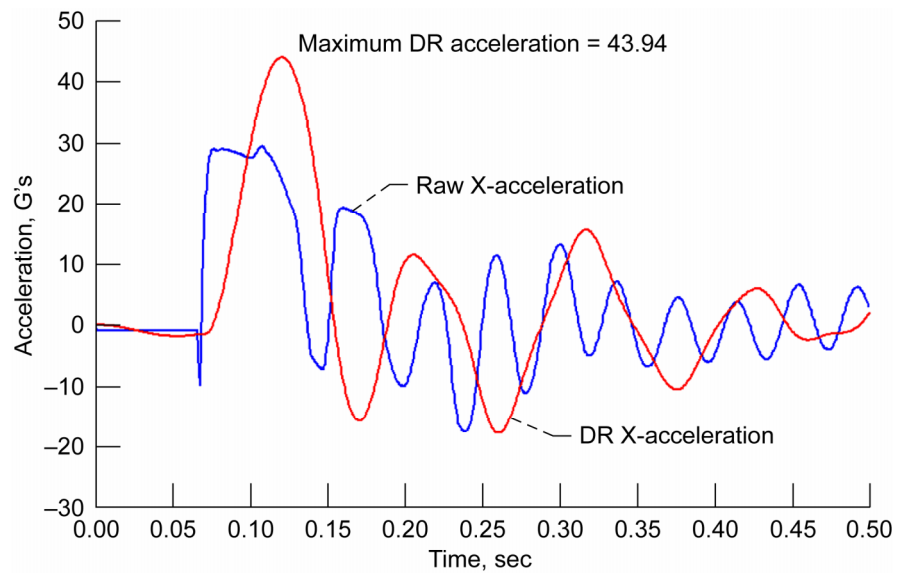


Figure 12.—Acceleration profile for high eyes in/out DR ($V_n = 58.33$ fps).

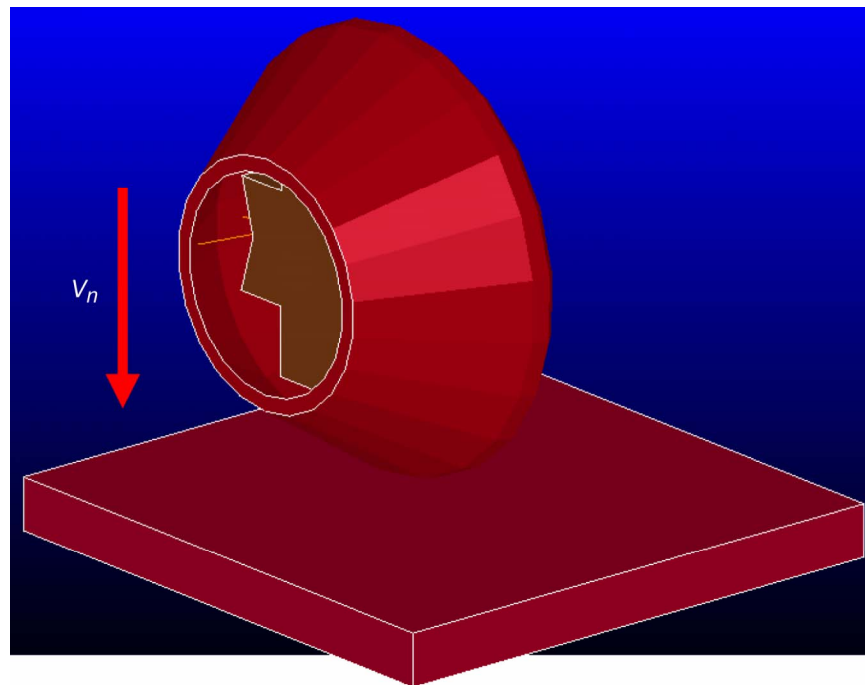


Figure 13.—Configuration for Y-acceleration (sideways DR).

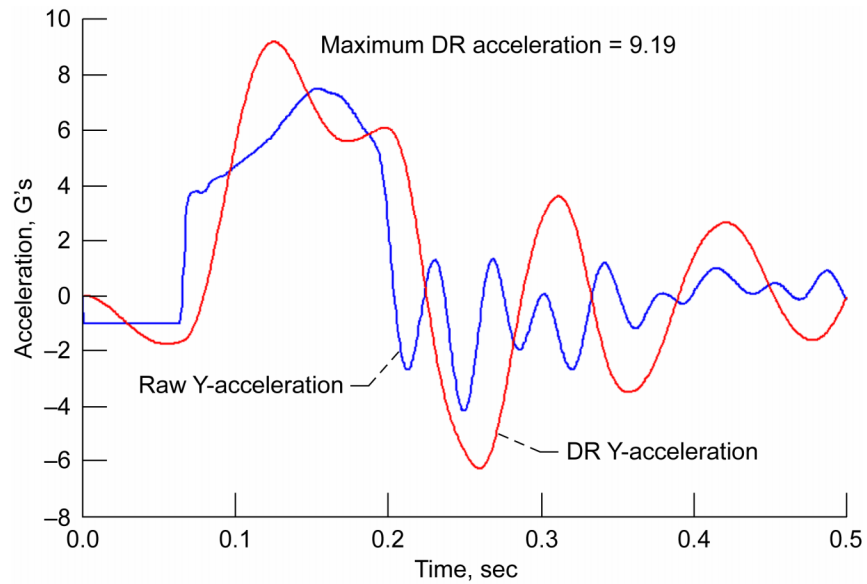


Figure 14.—Acceleration profile for low sideways DR ($V_n = 16.66$ fps).

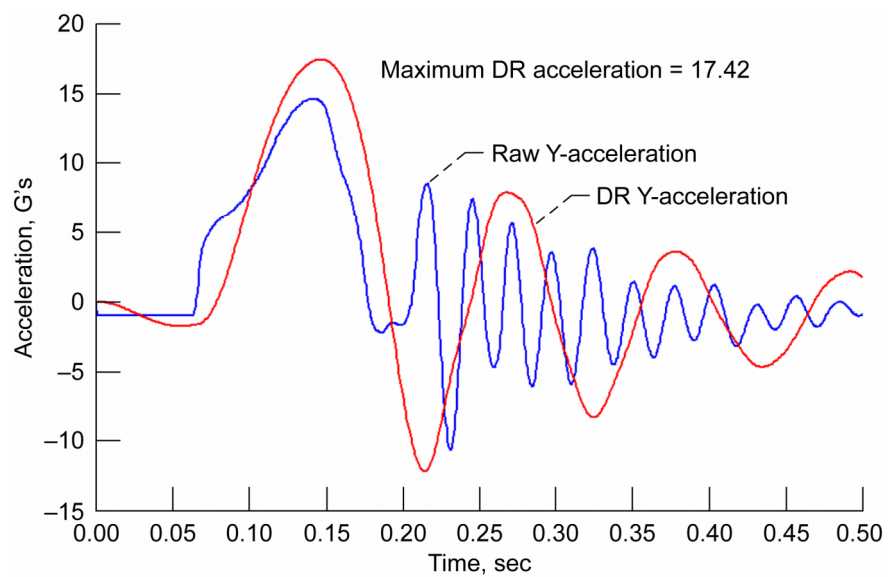


Figure 15.—Acceleration profile for moderate sideways DR ($V_n = 25$ fps).

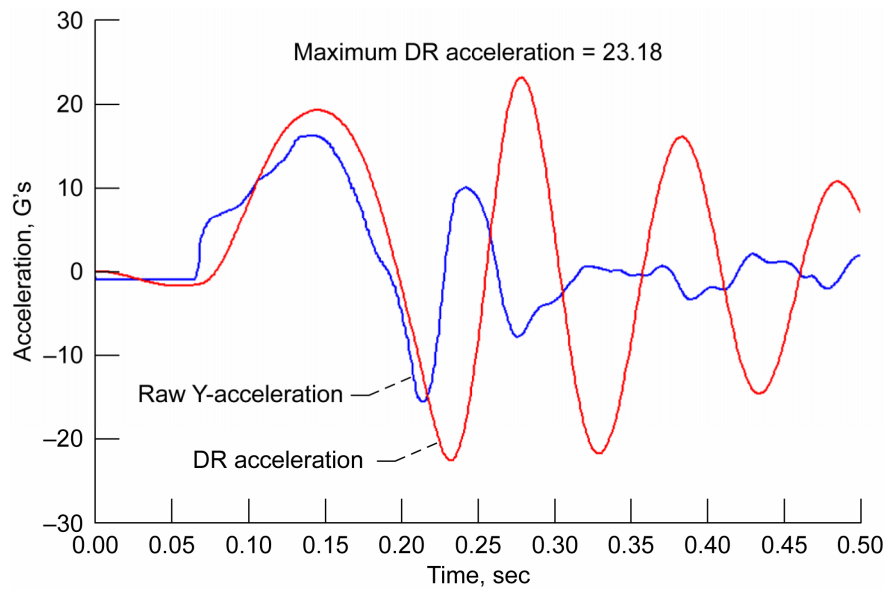


Figure 16.—Acceleration profile for high sideways DR ($V_n = 33.3$ fps).

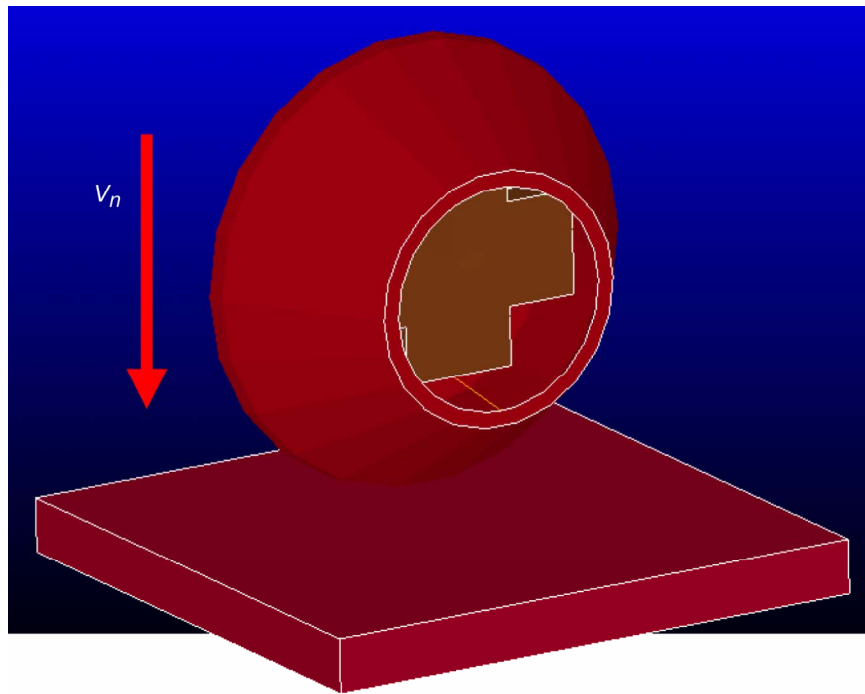


Figure 17.—Configuration for Z-acceleration (spine DR).

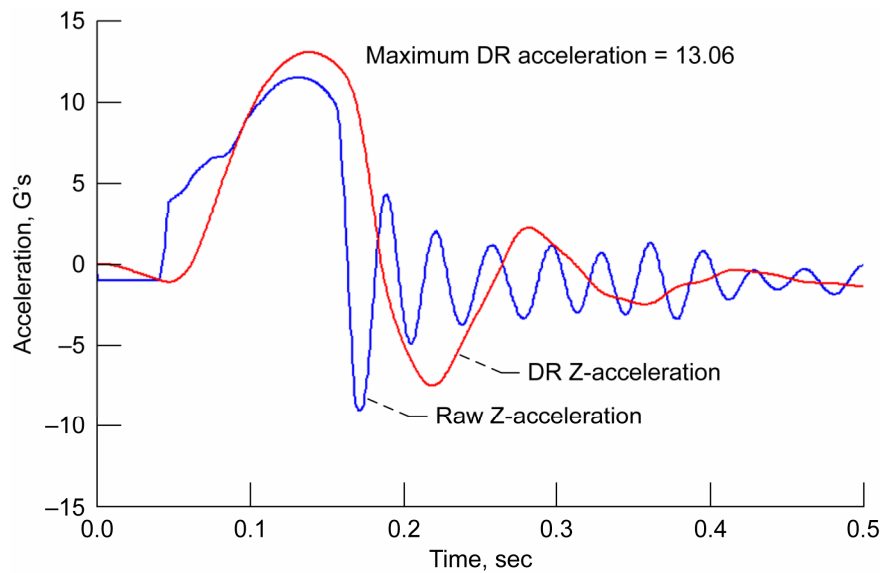


Figure 18.—Acceleration profile for low spine DR ($V_n = 25$ fps).

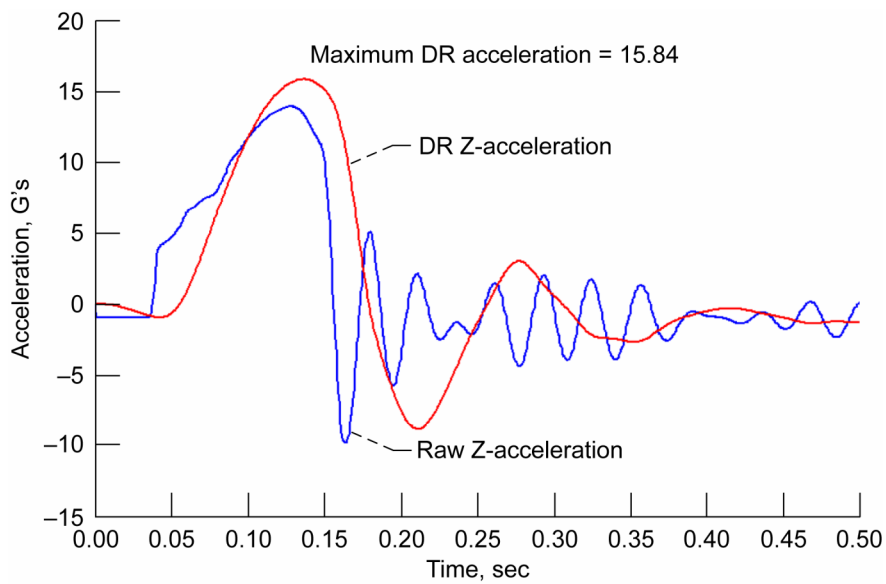


Figure 19.—Acceleration profile for moderate spine DR ($V_n = 29.2$ fps).

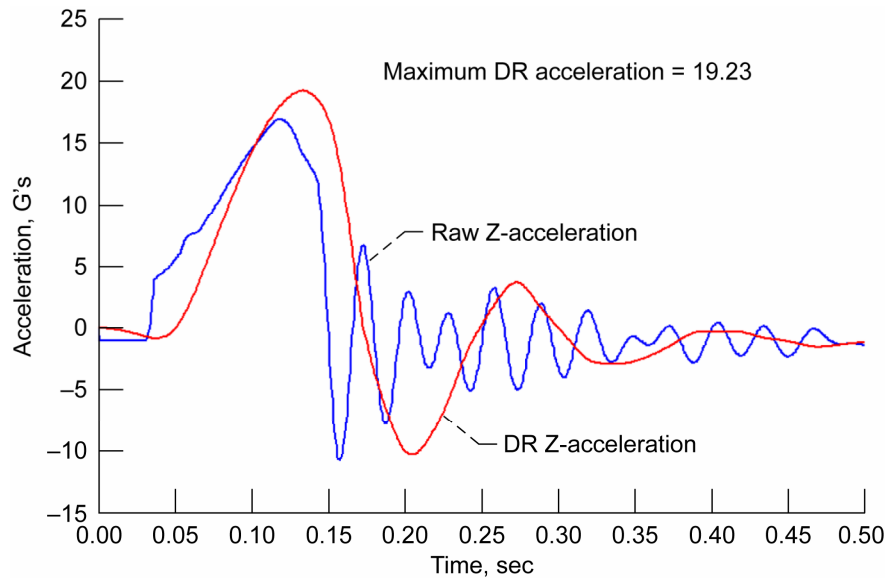


Figure 20.—Acceleration profile for high spine DR ($V_n = 33.3$ fps).

The FE crash test dummy provides a far greater amount of information concerning human body response including actual motions of the body such as limb flailing and head motion as well as loads within the body such as acceleration levels at specific locations in the body and forces on individual body parts. Figures 21 to 28 show details of the Hybrid III model and some of the locations in the dummy model where data is extracted. Figures 21 to 23 show the location of where accelerations were extracted in the head, pelvis and thorax. Accelerations at these locations are important since considerable research has been performed and injury criteria have been developed that provide correlations between acceleration levels and likelihood of injury. Figures 24 and 25 depict the lower lumbar region of the FE crash dummy model and where lumbar forces and moments are extracted while Figures 26 to 28 depict the neck region and locations where neck forces and moments are extracted. Similarly to the crash dummy accelerations, industry standards have been established that minimize injuries by limiting allowable predicted forces and moments in the crash test dummy. As previously mentioned in this report, it is important to note that many of the industry accepted standards have been developed for the automotive industry, or for other applications such as ejection seats, and may not be directly applicable to the type of conditions and injuries expected for the Orion crew members without further study and modification. It is also important to note that the FE test dummy is an analytical model of the physical dummy which is made up of non-human parts made from materials such as steel, aluminum, rubber and plastics. The FE test dummy is designed to predict the response of the physical test dummy and does not directly predict how a human might respond during an actual impact or the forces generated within the human body.

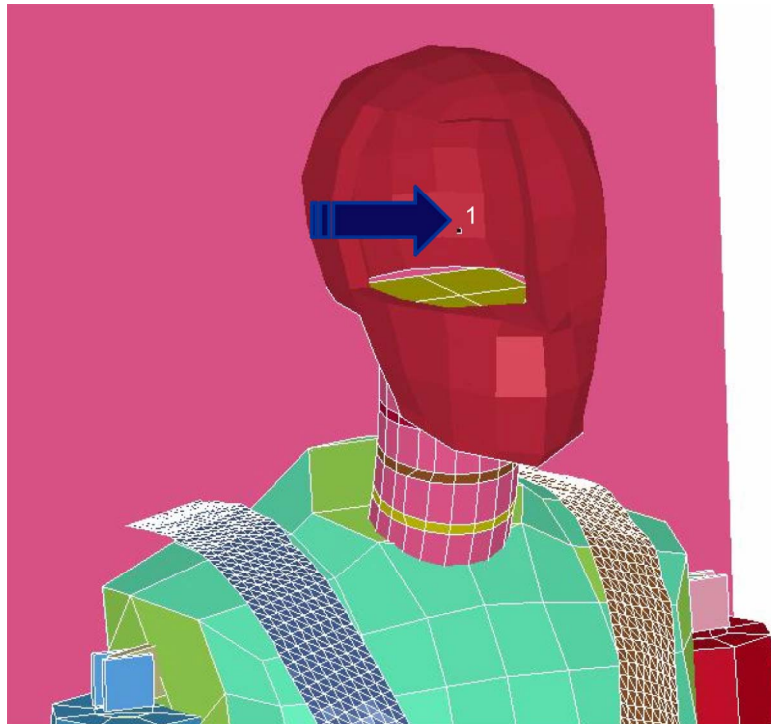


Figure 21.—Location of head acceleration.

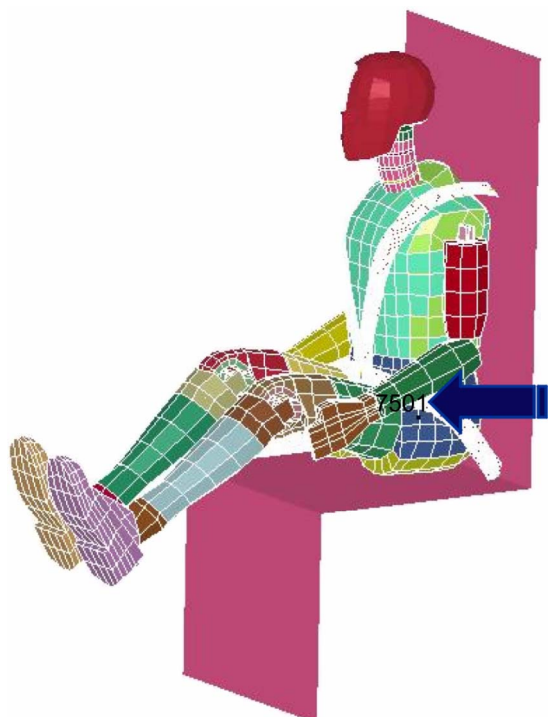


Figure 22.—Location of pelvis acceleration.

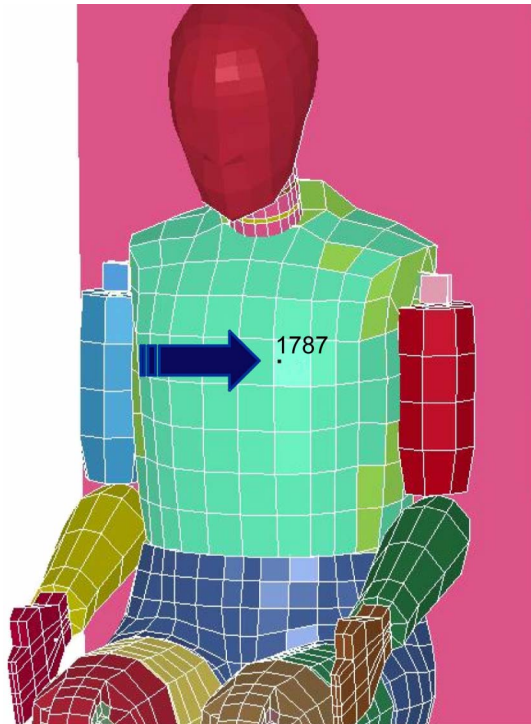


Figure 23.—Location of thorax acceleration.

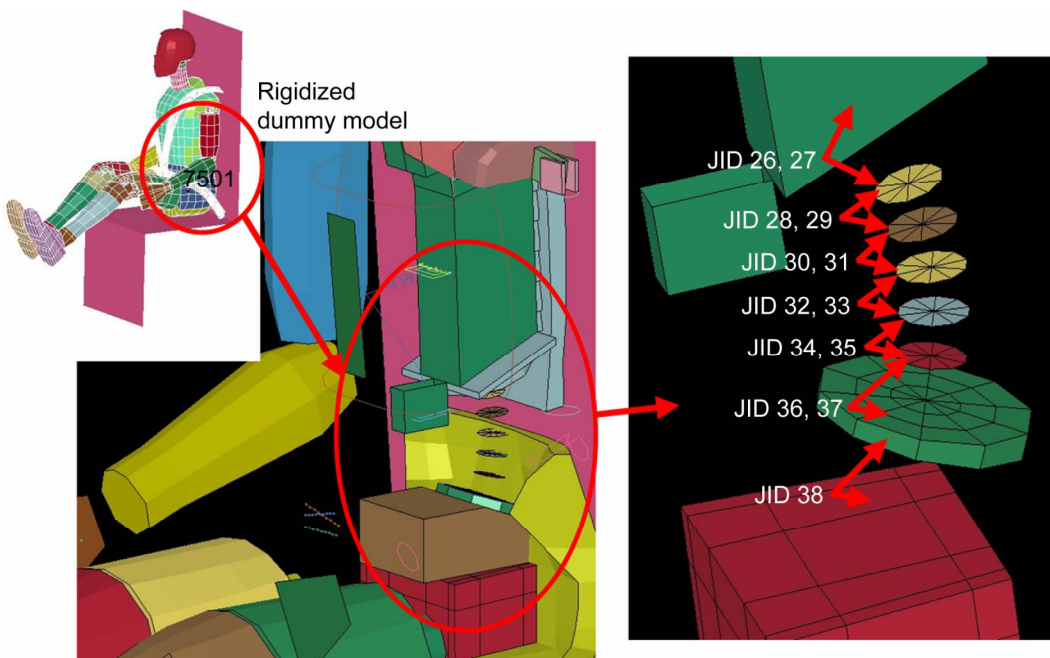


Figure 24.—Location of lower lumbar spine force.

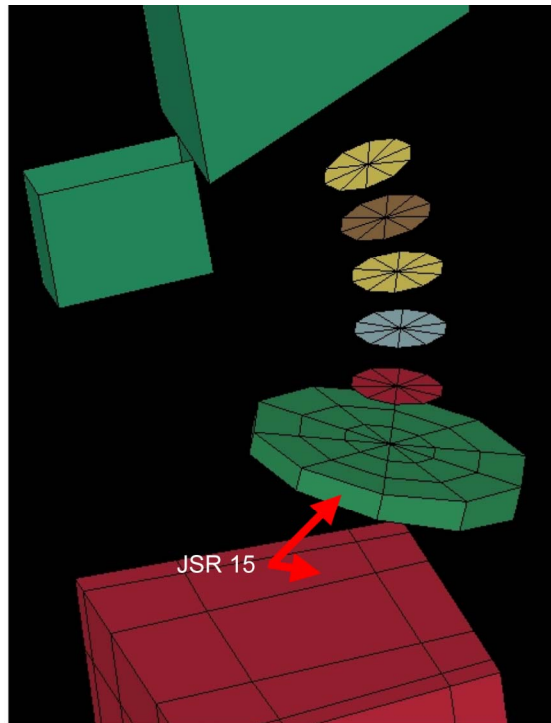


Figure 25.—Location of lower lumbar spine moment.

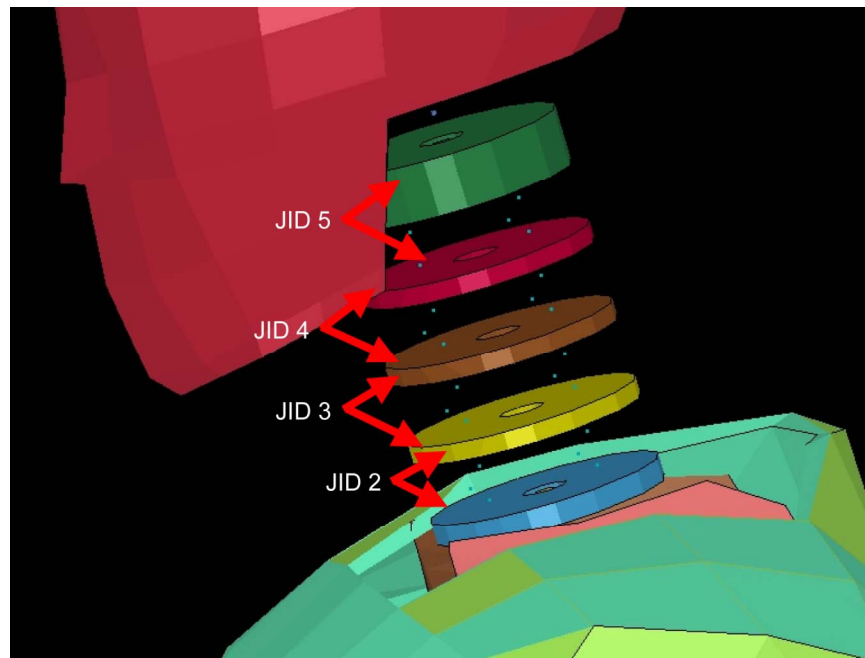


Figure 26.—Location of neck joints.

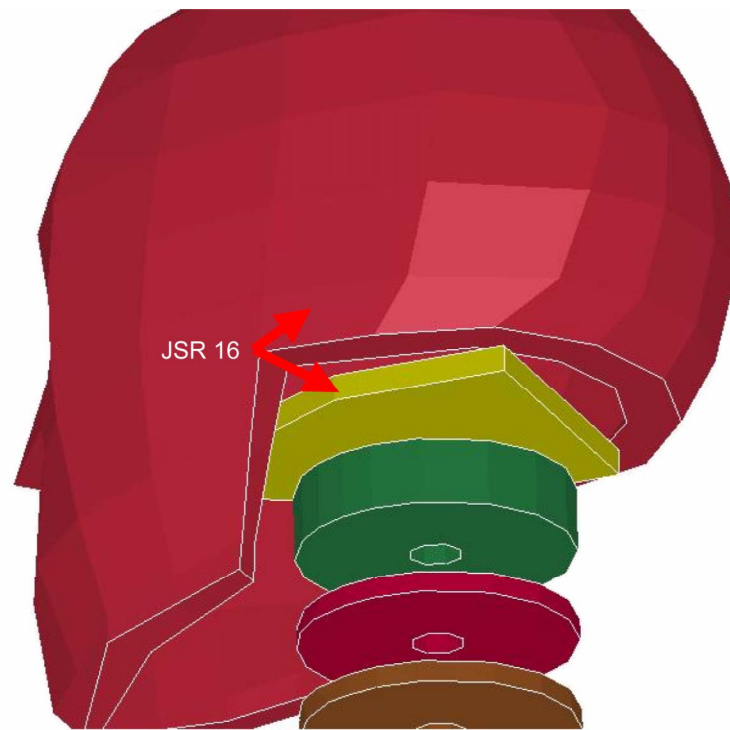


Figure 27.—Location of upper neck moment.

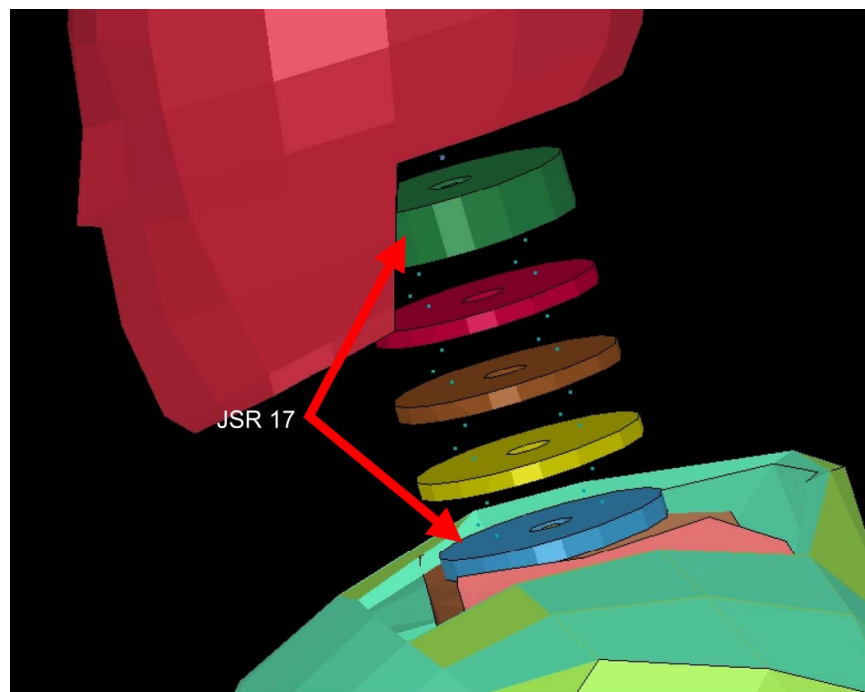


Figure 28.—Location of lower neck moment.

Z-Axis (Spinal) Acceleration Simulations and Results

Loading in the Z-axis (spinal) direction was examined since this is the loading direction that the Brinkley model was originally designed to address and where the Brinkley model is most accurate. Since the results of the Brinkley model are well anchored in actual data for loading in the spinal direction, it was useful to compare the Brinkley predictions to the results generated from the FE crash test dummy and to industry accepted injury criteria. Figure 29 shows the FE crash test dummy in a five-point harness. This harness was used since the five-point harness is the harness used for the development of the Brinkley model. The harness was composed of 2 in. nylon webbing tensioned to 50 lb in each strap. It is recognized that 50 lb is considered high and may be difficult to tolerate for more than brief periods and future work may use lower harness preloads. The harness was attached to a relatively rigid steel simplified seat and base accelerations were applied to the seat in the Z-axis direction (from the pelvis upwards towards the head) The commercially available Hybrid III FE crash test dummy model and LS-Dyna were used to generate the transient response.

Figure 30 and table 4 provide representative output from the LS-Dyna simulations. Figure 30 shows two intermediate time steps from the simulation. The first frame shows the maximum head flexion, which occurs just after the impact with the ground and the second frame shows maximum head extension, which occurs during the time period where the vehicle rebounds off of the ground. Table 4 provides a comparison among the three Brinkley injury levels and a collection of injury criteria from reference 1. Allowable limits for the injury criteria are included in the table and conditions where the limits are exceeded are identified by red. The Brinkley injury risk criteria are consistent for low levels of injury in that none of the injury criteria allowable are exceeded. For medium Brinkley injury risk the allowable thorax acceleration limit is exceeded while the remainder of the injury criteria is within allowable limits. For high injury risk for the Brinkley model, the thorax acceleration is close to double the allowable limit and the neck moment extension allowable limit is exceeded. It is not surprising that the allowable neck moment is exceeded considering the head is not constrained and undergoes considerable motion during impact. It is interesting that the lumbar force is within allowable limits for all three Brinkley injury levels

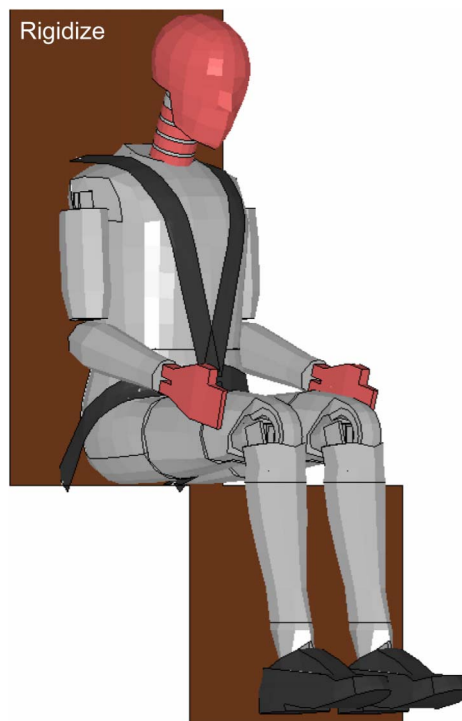


Figure 29.—Hybrid III crash test dummy constrained with five point harness.

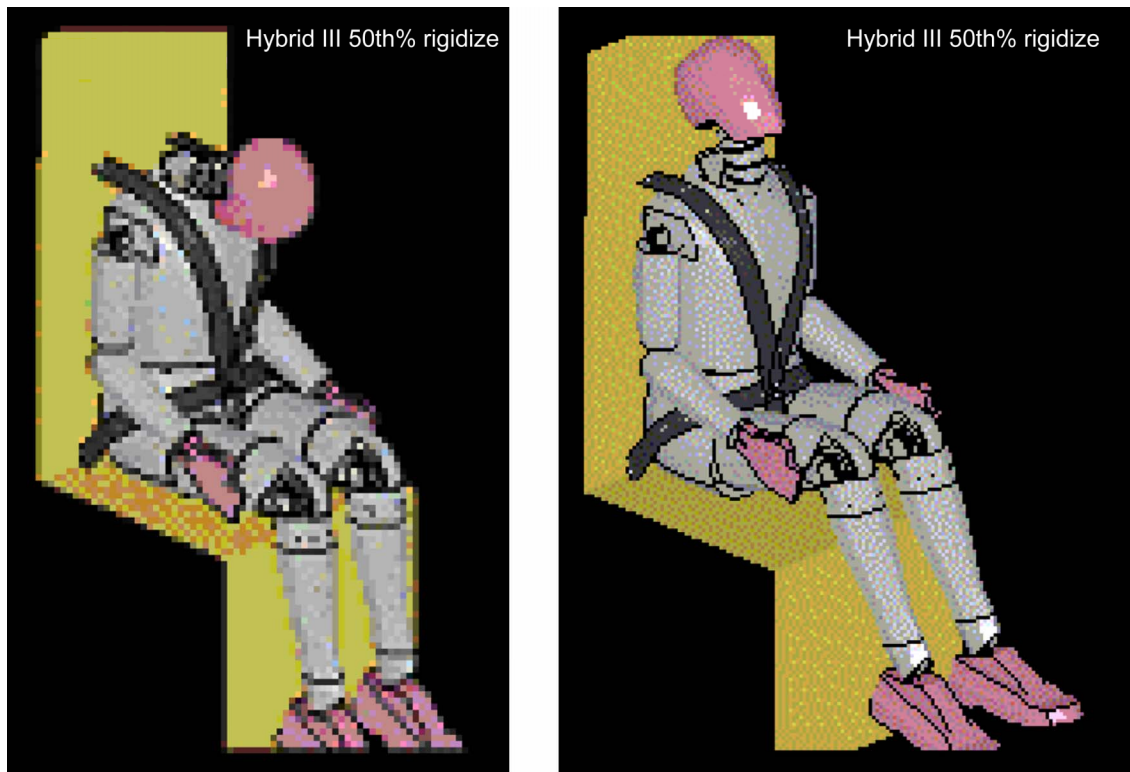


Figure 30.—Hybrid III crash test dummy response at intermediate time steps.

TABLE 4.—HYBRID III RESPONSE AND BRINKLEY INJURY LEVELS Z AXIS (SPINAL) EXCITATION

Criteria ¹	Location	Allowable	Brinkley low injury	Brinkley medium injury	Brinkley high injury
Head injury criteria (HIC)	N# 1 HIC	700	3.4	4.57	59.54
Chest severity index (CSI)	N# 1787 CSI	700	31.6	43	64.49
Thorax G's	N# 1787—resultant	60 G's	46.17 G's	75.27 G's	127.7 G's
Pelvis G's	N# 7501—resultant	-----	32.7 G's	49.42	41.9 G's
Lumbar force	JT# 26-37—resultant	6672 N	2804 N	4037 N	4545 N
	JT# 38—resultant	6672 N	4080 N	4500 N	5500 N
Lumbar moment	Jsr# 15—	-----	131.4 Nm	143 Nm	124.3 Nm
Neck force	JT# 2-5—resultant	6806 N	887 N	698 N	1277 N
Neck moment flexion (-)	JSR# 16—	310 Nm	66.42 Nm	78.6 Nm	79.42 Nm
Neck moment extension (+)	JSR# 17—	135 Nm	108.6 Nm	130 Nm	184 Nm

¹Trauma Biomechanics: Introduction to Accidental Injury, K.U. Schmitt, P. Niederer, F. Walz, Springer 2004.

considering that for this direction of loading spinal cord injuries are expected to be most prevalent (Bowman, 1993). It is important to note that many of the industry accepted standards have been developed for the automotive industry, or for other applications such as ejection seats, and may not be directly applicable to the type of conditions and injuries expected for the Orion crew members. For example, many of the automotive injury criteria were developed for frontal impacts where the human body rapidly decelerates and body parts such as the head impacts some part of the automobile interior. Similarly, aircraft ejection seat criteria are designed for an upright seated pilot being ejected vertically through the aircraft canopy where the primary loading is up through the spine. For Orion nominal landings, the crew members will be seated on their backs and the primary loading will be in a direction from the crew member's back towards the chest. Further study is recommended before existing industry accepted injury criteria developed for applications such as automotive or military are used to assess Orion crew member injury.

It is important to note that the probabilities for injury are different for the Brinkley criteria and the allowables used to assess the HYBRID III results. For example, a Brinkley medium and high injury corresponds to a probability of serious injury of 5 and 50 percent respectively, while exceeding the thorax 60 G limit corresponds to a 25 percent probability of exceeding a 4 on the Abbreviated Injury Scale (AIS) where 4 is indicative of severe injury. To further complicate matters, the HYBRID III dummy response allowables are often not consistent with each other. For example, the head injury criteria limit of 700 corresponds to a 31 percent probability of moderate skull fracture while the thorax limit of 60 g's corresponds to a 25 percent probability of severe injury. In this case neither the probability nor the severity of injury is the same.

Y-Axis (Sideways) Acceleration Simulations and Results

Results of FEA simulations for the standard five-point harness were compared with simulations that contained a variety of lateral supports inspired by the example of a modern racecar seat (fig. 31). Modern racecar seats provide excellent driver protection during impacts that might normally be fatal with a more traditional seat design (Gramling, H. et al., Melvin, J.W et al.). These seats provide a higher level of protection through the use of lateral supports, and head and neck restraint systems, as well as improved harnesses. A typical modern race seat is shown in figure 31 along side the seats from the Apollo program. The differences in lateral support between the two seat designs are clear in the images. Much of the improvements that have been realized in racecar seats are the results of trial and error and results obtained during actual operations and accidents. Orion seat designs cannot be designed via trial and error or in-use test programs and therefore to take advantage of any improvements in restraint or seat designs physical and FE crash test dummies will need to be used.

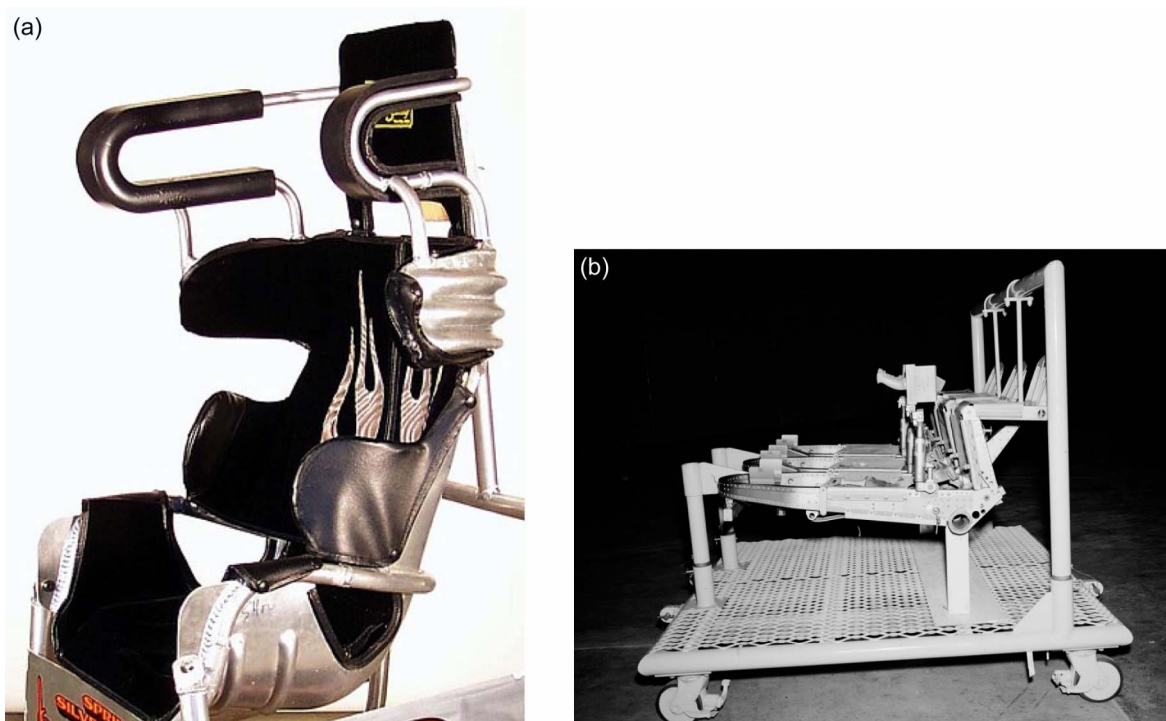


Figure 31.—Comparison between modern racecar and Apollo seats. (a) Modern racecar seat. (b) Apollo seat (crew member seated on back).

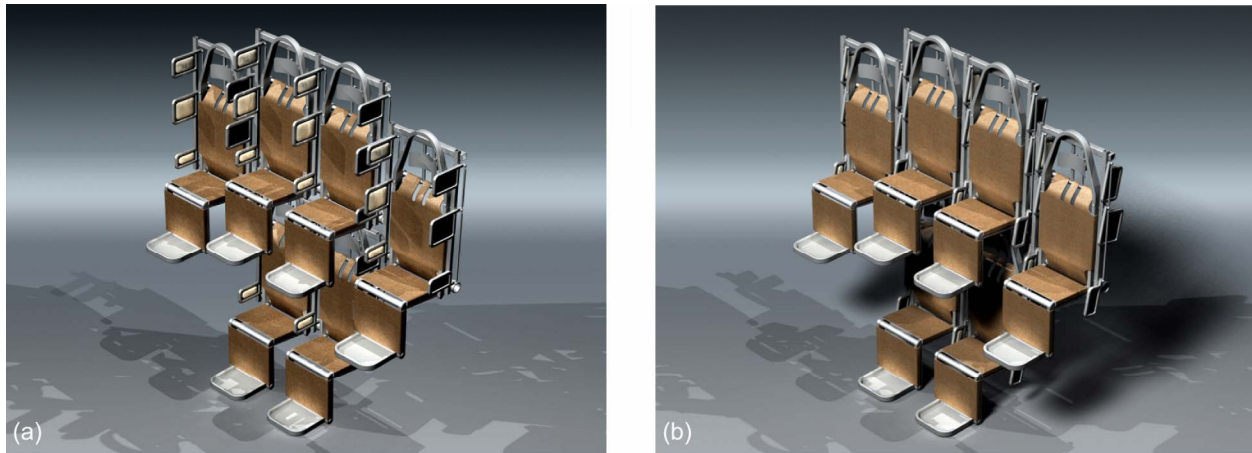


Figure 32.—Airbag concept shown in a six-seat configuration. (a) Airbag system deployed. (b) Airbag system stowed.

The use of lateral supports of the type found in racecar seats may not be directly applicable to Orion since these seats may be impractical for ingress and egress of the crew and for stowage during flight. In order for lateral restraints (or any other restraint or crew protection system) to be used in space flight applications the system must provide for a practical means for ingress, and rapid egress, and stowage when the system is not in use. In addition the restraint system must be flexible in its ability to accommodate crewmembers of different sizes and weights. One system that shows particular promise is a series of side airbags mounted on panels that are linked together and can be quickly extended and retracted by the crew for ingress and egress (fig. 32). In this concept there is a retractable arrangement of airbags on each side of the individual seats. The airbags are mounted so that they pivot about one corner, and there is a mechanical linkage that connects them so that they move together as a group. In the up position they are held by a spring clip and by manually pushing any one of the airbags down the entire group can be quickly stowed. The airbag position can be changed for taller or shorter crewmembers by using an adjustable length linkage between airbag pallets.

Simulations for impacts in the direction of the Y-axis were conducted using the same Hybrid III FE model, harness and seat as was used for the Z-axis study in the previous section. The purpose of performing simulations in the sideways direction was to evaluate the effect of various combinations of side impact constraint devices on crewmember injury and to assess the general effectiveness of the FE crash test dummy for performing this type of study. The Hybrid III FE crash test dummy is not ideally suited for simulating side impacts; however, since this model was already generated for the work performed in the previous section it was expedient to re-use the model for side impacts. The results presented in this section may not be as accurate as if they had been computed with a better-suited dummy such as the Side Impact Dummy (SID) however, for performing a first order comparison of the effect of side constraints this model was adequate.

Figure 33 shows the four of the five configurations that were used to assess the effectiveness of side impact constraints. The baseline configuration was an unconstrained crewmember in the five-point harness. The second configuration was comprised of thin pads with a three-inch gap between the crewmember's head and shoulders. The third configuration (not shown in figure) was thicker pads with a one-inch gap between the crewmember's head and shoulder. The fourth configuration was comprised of thick pads in direct contact with the crew members head and shoulders and the fifth configuration was identical to the fourth with thick pads and hand and feet constraints were added to retain the crew member's arms and legs from significant flaying. The properties and thickness of the various pads are important for detailed analysis however for this preliminary study the pads were not optimized and the results are only indicative of overall trends. All of the simulations performed in this section of the report were done using the input acceleration profile that generated a high injury risk using the Brinkley model in the sideways direction.

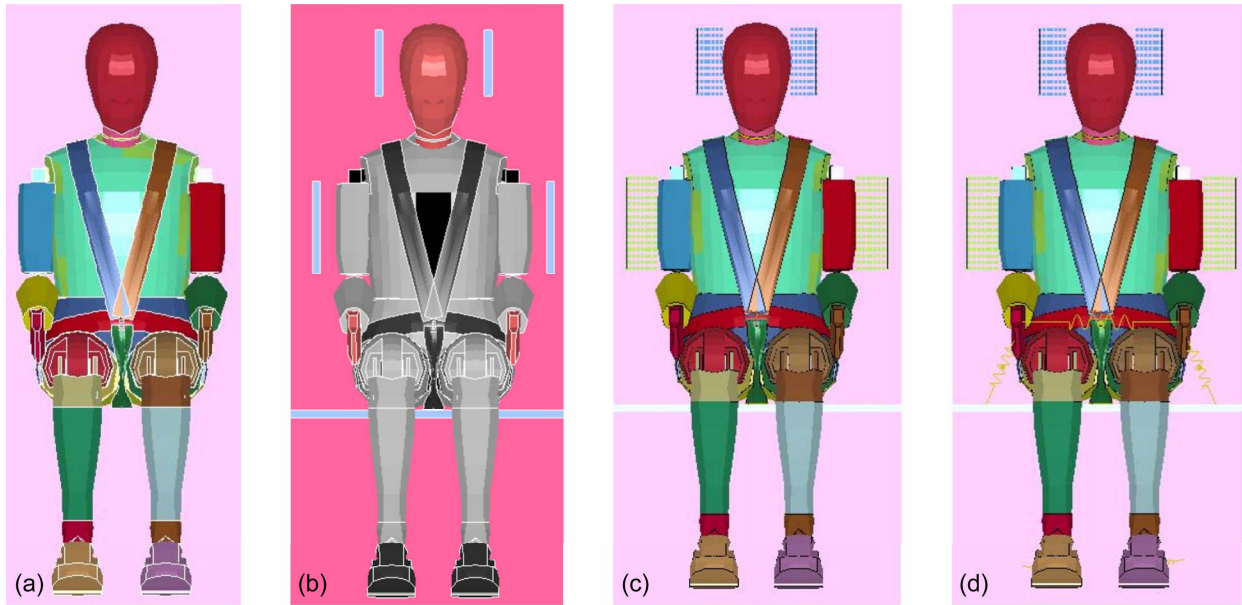


Figure 33.—Comparison among padded and constrained crew member for sideways loading. (a) Unconstrained. (b) Thin pads. (c) Thick pads. (d) Thick pads and hands and feet tethered.

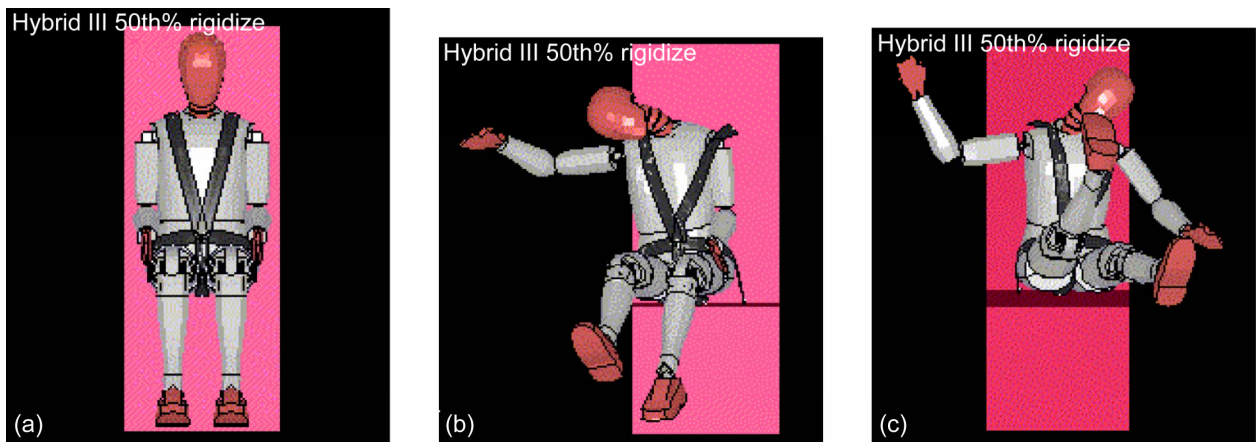


Figure 34.—Hybrid III crash test dummy response at intermediate time steps for side impact (no side padding). (a) Initial conditions. (b) Initial impact. (c) Rebound.

Figure 34 and table 5 show the results for the first configuration where the crewmember is only constrained in the five-point harness. Figure 34 shows the crash test dummy response at three intermediate time steps for the model without any side padding. Large excursions of the head, arms and legs are shown clearly during both the initial impact and the rebound. The results in table 5 agree with the Brinkley prediction of high injury risk considering that four of the seven injury criteria are exceeded including the Chest Severity Index (CSI), the thorax maximum allowable G and the neck flexion and extension. It is important to point out that the neck injury criteria are intended for frontal impacts and have been very loosely applied to the present application of side impact. For side impact there are no clearly defined neck moment criteria however it was felt that the frontal impact neck criteria could be reasonably applied, as an approximation, to side impacts. Additional study of neck injury criteria is certainly required before firm criteria can be established.

TABLE 5.—COMPARISON AMONG DUMMY CONFIGURATIONS USING INPUT ACCELERATION PROFILE THAT PRODUCES HIGH BRINKLEY INJURY IN LATERAL (SIDEWAYS) DIRECTION

	Allowable	Five point harness*	Thin pads	Thicker pads, 1 in. gap	Thick pads	Thick pads, hand and feet constraints
HIC (Node 1)	700	225	4050	256	174	92
CSI (Node 1787)	700	965	1960	1965	1211	1559
Thorax G's (Node 1787 filtered @ 180 Hz)	60 G's	78 G's	90 G's	62 G's	71 G's	81 G's
Pelvis G's (Node 7501 filtered @ 180 Hz)	130 G's	90 G's	120 G's	90 G's	161 G's	126 G's
Neck force (Max JT2-5 filtered @ 180 Hz)	6806 N	3950 N	3500 N	3950 N	3264 N	3283 N
Neck moment flexion (Max +JSR17 ϕ, θ, ψ , filtered @ 180 Hz)	310 Nm**	315 Nm	165 Nm	142 Nm	123 Nm	84 Nm
Neck moment extension (Max +JSR17 ϕ, θ, ψ , filtered @ 180 Hz)	135 Nm**	180 Nm	130 Nm	125 Nm	132 Nm	96 Nm

*All configurations utilize five point harness

**Frontal impact criteria used for lateral direction

Figure 35 and table 5 show the results for the second configuration where the crewmember is constrained in the five-point harness and thin side padding is placed three inches away from the head and shoulders. This type of support is similar, in concept, to what is currently used for racecar seat side supports. This type of support has been proven highly successful in reducing injury during relatively violent high-speed racecar crashes and was thought to have similar benefits for Orion crew members also undergoing side impact loadings. Figure 35 shows the crash test dummy response at three intermediate time steps for the model with thin side padding. The large excursions of the head, arms and legs that were seen for the configuration without side padding is eliminated with the thin pad configuration. The results in table 5 show that the thin side pads help to reduce the neck moments, which was expected since the side pads prevent the head from bending over. However, the Head Injury Criteria (HIC), CSI and thorax and pelvis acceleration criteria all exceed their allowable limits and are considerably larger than for the configuration without any pads. These results were also expected since the crewmember head and chest impact into the relatively thin and stiff pads leading to short durations of very large accelerations.

The configuration with the thicker pads and 1-in. gap between the pads and head and chest provides a higher level of injury protection than the previous two cases. For this configuration, only the chest severity index and the thorax maximum allowable acceleration are exceeded. In fact, the thicker pads provide so much reduction in head acceleration that the head injury criterion is reduced by an order of magnitude over the configuration with thin pads. For the thick pad configuration where the pads are in contact with the head and shoulders, the head injury criteria and the chest severity index are further reduced however; the pelvis acceleration is increased and is beyond the allowable limit. For the thick pads and constrained arm and leg configuration, all of the observed injury criteria are reduced compared to the other configurations except for the chest severity index and the thorax acceleration. Both of these criteria exceed their allowable limits.

In general, introduction of the side supports eliminates the large excursion of the head and the hyperextension of the neck. Neck hyperextensions are the main cause of severe injury for the case without lateral support, so it appears promising that lateral support will improve survivability for Y-axis direction impact. However the body must still mitigate the crash, so while neck forces are reduced due to reduced head and neck extension, other body forces and moments may show increases. This is supported by the fact that padding had the effect of reducing the neck moment while increasing the chest severity index. The key design goal is to keep any individual force, moment or acceleration from exceeding the allowable injury level, which will require tradeoffs in the design. With further design improvements it may be

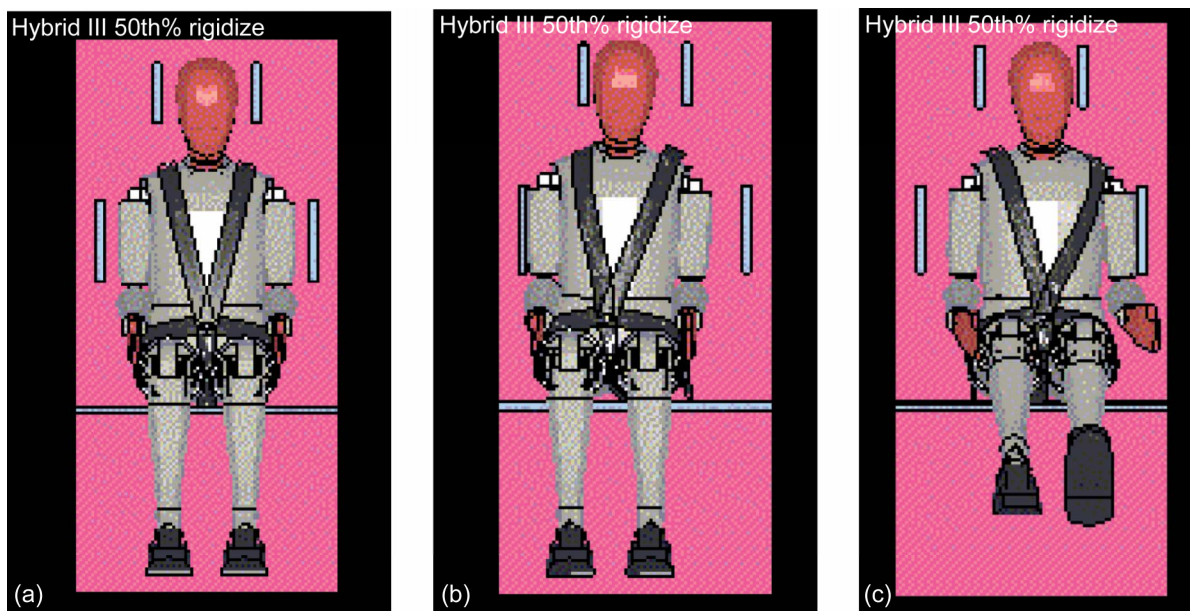


Figure 35.—Hybrid III crash test dummy response at intermediate time steps for side impact (thin side padding).
(a) Initial conditions. (b) Initial impact. (c) Rebound.

possible to reduce the chest and thorax loadings to within allowable limits, however, for the designs that were considered, both the Brinkley model and the FE crash test dummy predicted high levels of injury.

A significant difference between the Brinkley model and the FE crash test dummy is that the Brinkley model cannot directly determine the effects of adding constraints such as side padding while the FE crash test dummy can simulate somewhat realistically the distributed motion of the body as well as predict accelerations and loads at discrete locations through the body. Furthermore, the Brinkley model only estimates the risk of injury, not the location or type of injury. The FE test crash test dummy provides a major advantage of the FE crash test dummy over the Brinkley model in that the dummy may be used to assess the effects of different seat designs, constraint systems, helmets and crew protection systems.

X-Axis (Chest In/Out) Acceleration Simulations and Results

A comparison between the Brinkley model and the FE crash test dummy was made in the X-axis direction using an applied acceleration profile obtained from a maximum drag abort situation. This situation occurs when an emergency occurs during ascent and the crew module must be separated from the launch vehicle using the Launch Abort System (LAS). When the LAS is activated, the crewmembers are pushed into the backs of their seats while the LAS rockets are fired. Once the LAS rocket's fuel is depleted they cease firing and aerodynamic drag reverses the acceleration on the crewmembers and they are pulled out of their seats in the opposite direction. The Hybrid III FE crash test dummy is probably most suited for loadings in the +X-axis direction since this is the direction where the test dummy was originally designed for automotive frontal impacts.

The acceleration profile for the maximum drag abort is shown in figure 36 and the corresponding FE crash test dummy response at intermediate time steps is shown in figure 37. As expected, during LAS rocket firing the crewmember is pushed into his seat and there is minimal motion of the head, arms or legs. In the later stage when aerodynamic drag is dominant and the crew member is pulled out of his seat, both the head and arms and legs are extended leading to the potential for larger neck forces and head accelerations. (Note: this particular model contained overly stiff elbow joints which explains the limited extension of the arms) Additionally, flailing of the arms and legs in the close confines of Orion could be an issue since the arms and legs may impact surrounding structures or strike other crewmembers. To

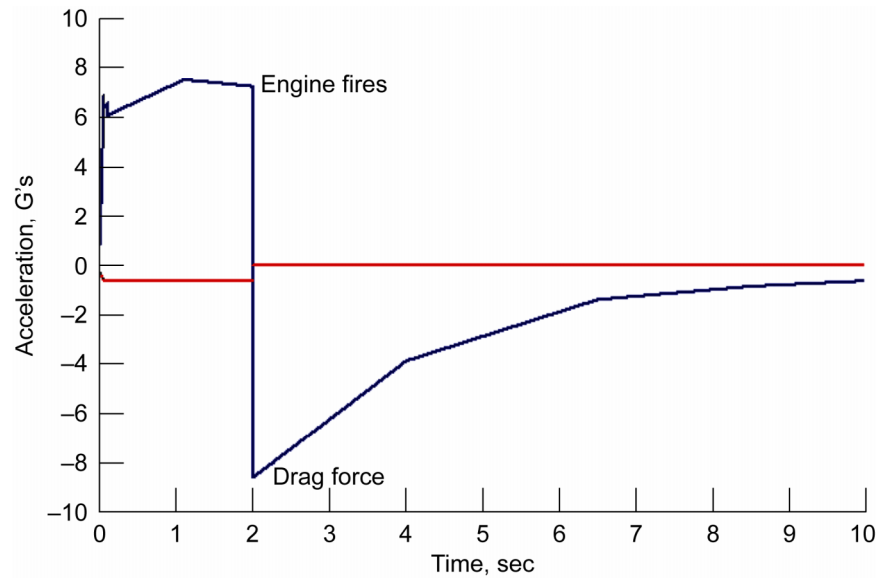


Figure 36.—Acceleration profile for maximum drag abort (from Constellation program loads data book, figure 2.7.1.3.1.4-2).

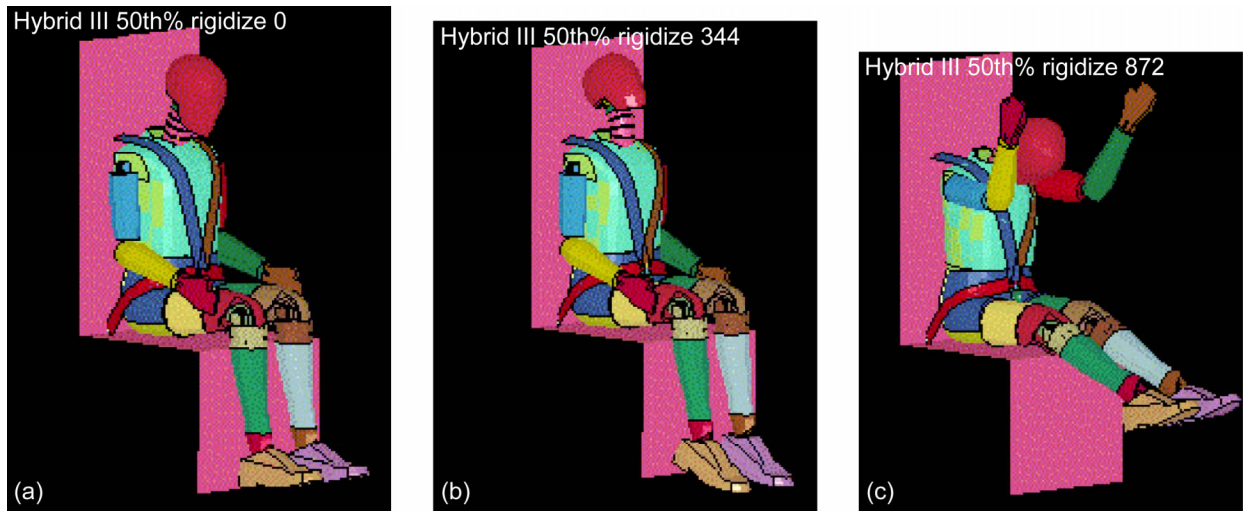


Figure 37.—Hybrid III crash test dummy response at intermediate time steps for maximum drag abort. (a) Initial conditions. (b) Large abort system rocket firing. (c) Aerodynamic drag.

compensate for flailing, hand and foot restraints will be important. The large head movement could also be an issue and some type of head/helmet constraint may be required.

Figure 38 shows the maximum drag abort acceleration that was applied to the crewmember seat along with the resulting Brinkley dynamic response and the FE crash test dummy response in the X-axis direction. Both the Brinkley and the dummy response follow, reasonably close, both the input acceleration shape and magnitude. The Brinkley response very closely tracks the input acceleration profile during the rocket-firing phase then oscillates about the input acceleration during the aerodynamic drag phase. The Brinkley model, as mentioned previously, employs an approximately 10 Hz oscillator, so it is expected that the response after the load reversal would oscillate and that the oscillations would be near 10 Hz. The maximum Brinkley dynamic response is 20 G, which is within the Brinkley low risk of

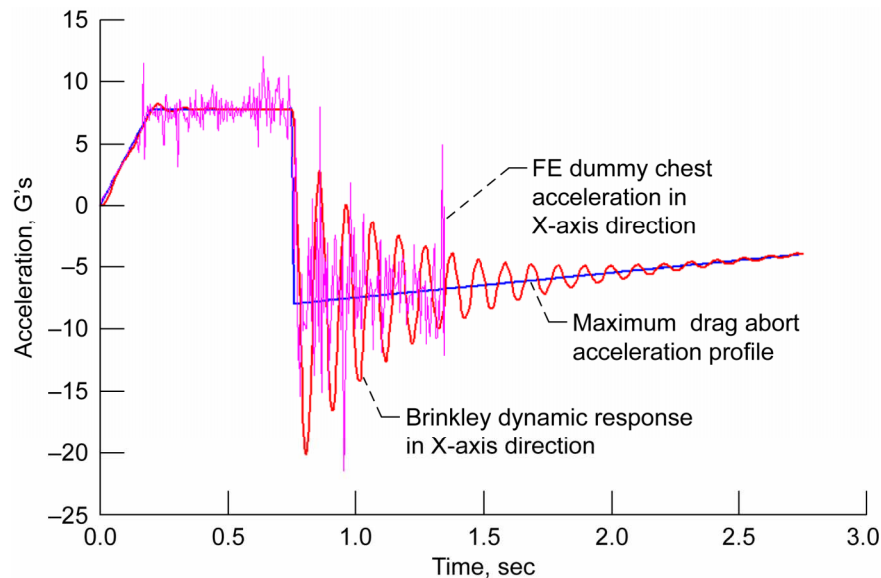


Figure 38.—Acceleration profile for maximum drag abort, Brinkley dynamic response and FE dummy chest acceleration.

injury range for X-axis (chest in/out) accelerations. The FE crash test dummy response is within the same general range as the Brinkley response except for numerous short duration acceleration spikes through the transient response. The dummy response has been filtered to eliminate response above 180 Hz and further reductions, or elimination, of at least some of these acceleration spikes could be eliminated by adding more damping to the dummy model and refinement of the model through the inclusion of a higher fidelity seat model and a small amount of seat padding. Even with these outlying acceleration peaks, the computed CSI is well below the allowable limit ($212 < 700$) leading to a consistent result with the Brinkley prediction of a low probability of injury.

Conclusions and Recommendations

The FE crash test dummies used in conjunction with the Brinkley model provides a useful set of tools for predicting potential crew injuries during vehicle landings. In addition these tools enable the design and evaluation of alternate crew protection systems. Specifically

1. The Brinkley model reasonably ensures an acceptable environment for crew members in a five point harness; however, it has limited ability to provide insight into additional or modified crew restraint and protection systems. The Brinkley Model has been modified to included the general effect of sideways crew member support, however, the Brinkley Model does not distinguish between types of support or their differences in effectiveness in preventing injury.
2. FE analysis and crash test dummy tests are capable of providing valuable insight into alternate crew member injury protection systems and should be employed for this purpose. The FEA model and the physical crash test dummies can be used to assess the effects of variations of restraint and support in a comparative manner and if validated with human response data, can provide quantitative assessments of crew injuries.
3. Additional evaluation of automobile industry safety standard injury criteria should be initiated to ascertain the applicability of these criteria to crew member protection.
4. Practical methods for implementing additional crew protection appears realistic and should be further developed.

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